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TITLE: Magnetic apparatus for MRI

Abstract Text (1):
Magnetic apparatus for MRI/MRT probes and methods for construction thereof are disclosed. One embodiment includes a pair of opposed magnet assemblies defining an open region therebetween, a transmitting RF coil having at least a portion thereof disposed within the open region, at least one receiving RF coil disposed within the open region and X,Y and Z gradient coils. At least one of the X,Y and Z gradient coils is disposed outside of the open region. Another embodiment of the apparatus includes a single magnet assembly having a first surface and a second surface opposing the first surface, a transmitting RF coil having at least a portion thereof opposing the first surface, at least one receiving RF coil and X,Y and Z gradient coils. At least one of the X,Y and Z gradient coils opposes the second surface. In another embodiment the magnet assembly generates a permanent z-gradient magnetic field and therefore includes only X and Y gradient coils, at least one of which opposes the second surface. The apparatuses may also include one or more shim coils.

Brief Summary Text (2):
The present invention is generally related to the fields of magnetic resonance imaging (MRI) and magnetic resonance therapy (MRT).

Brief Summary Text (4):
MRI systems for performing whole body imaging usually employ large magnets which effectively surround the patient. Such magnets are usually large superconductor magnets which are expensive and difficult to maintain. MRI systems for performing local imaging of specific body parts or organs are known in the art.

Brief Summary Text (5):
U.S. patent application Ser. No. 08/898,773, now U.S. Pat. No. 5,900,793 to Katznelson et al., filed Jul. 23, 1997 and entitled "PERMANENT MAGNET ASSEMBLIES FOR USE IN MEDICAL APPLICATIONS", now U.S. Pat. No. 5,900,793 and incorporated herein by reference discloses, inter alia, compact permanent magnet assemblies for use in medical applications including MRI and/or MRT.

Brief Summary Text (6):
A typical application using an intra-operative MRI system is brain surgery. Reference is now made to FIG. 1 which is a schematic perspective view of a small organ dedicated MRI probe useful in brain surgery. The MRI probe 1 includes two annular permanent magnet assemblies 2 and 4 connected by a frame 3. The frame 3 and the magnet assemblies 2 and 4 are shaped for imaging the brain of a patient 6. During MRI assisted brain surgery or MRT, the head of the patient 6 is positioned between the two magnet assemblies 2 and 4.

Brief Summary Text (11):
The MRI probe 1 further includes Gradient coils (not shown) for generating gradient fields, shim coils (not shown) for active shimming of the main magnetic field, RF coils (not shown), a temperature control system (not shown) and an RF shield (not shown).

Brief Summary Text (12):
Ordinarily, the gradient fields are generated by a set of coils, through which a current of an adequate magnitude flows. During the periods of building up and decay of the currents, the temporal change of the magnetic flux, originally generated by

the currents, created eddy currents in conductive materials situated in their vicinity such as soft iron parts or permanent magnet parts used in prior art MRI permanent magnets or the aluminum enclosures of the cooling systems used in super-conducting magnets of MRI systems. The eddy currents generated by the gradient coil magnetic flux changes, generate secondary magnetic fields which may interfere with the primary gradient fields and affect their precision in encoding the spatial information.

Brief Summary Text (13):
In prior art MRI devices, the gradient coils are located within the internal free volume situated in the main magnet, where the imaged body is also introduced. To attenuate the effect of the spurious eddy currents, prior art MRI devices may use shielded gradient coils or pre-emphasis circuits which modify gradient amplifier demand in order to compensate for eddy currents. In small organ dedicated MRI probes and in MRI probes adapted for intra-operative use such as the MRI probe 1 of FIG. 1, the dimensions of the region 16 (best seen in FIG. 2) for accommodating the organ to be imaged are limited by practical considerations. Generally, the design of such MRI systems involves a tradeoff between maximizing the intensity and homogeneity of the magnetic field in as large an imaging volume as possible and providing maximal accessibility of the surgeon to the organ undergoing surgery. For example, the MRI probe 1 (FIGS. 1 and 2) is designed to maximize the size of the volume 18 of homogenous magnetic field while keeping the size of the magnet assemblies 2 and 4 minimal while allowing enough space for positioning the shoulders of the patient 6. If one tries to increase the space available for the shoulders of the patient 6 by increasing the distance between the magnet assemblies 2 and 4 along the axis 12, the resulting decrease in the strength and homogeneity of the magnetic field will have to be compensated. The magnetic field can be compensated by increasing the thickness of the annular permanent magnets 4a, 4b, 4c of FIG. 2 and 2a, 2b and 2c (not shown in FIG. 2).

Brief Summary Text (17):
Furthermore, in MRI systems using permanent magnets, if the gradient coils are positioned in close proximity to the permanent magnets, the heat developed in the resistive gradient coils by the currents flowing within the coils may heat the permanent magnet. The heat generated by the gradient coils may thus cause local temperature increase in the permanent magnets. Such temperature changes are undesirable since the field generated by permanent magnets is highly susceptible to large variations induced by local temperature changes.

Brief Summary Text (18):
MRI systems based on permanent magnets such as the MRI probe 1 of FIG. 1 or the MRI probe of FIG. 2, do not include electrically conducting structures operating as magnetic flux return structures. This fact, in addition to the segmented structure of the annular permanent magnets 4a, 4b, 4c and 2a, 2b and 2c (not shown) and the intrinsic low conductivity of the Nd--Fe--B alloy from which they are made, substantially reduce the spurious eddy current problem.

Brief Summary Text (19):
Whole body MRI/MRT systems typically use a fixed installation RF cage for preventing magnetic, electromagnetic and electrical noise from the outside from penetrating into the imaging volume inside the probe and interfering with the weak NMR signals generated during imaging. In addition, the RF cage is also used to reduce the leakage of the RF radiation generated within the probe during imaging to prevent disturbances to other electrical devices used near the MRI probe.

Brief Summary Text (20):
Unfortunately, for practical reasons, large fixed installation RF cages or RF rooms cannot always be used small organ dedicated MRI or MRT probes of the type used for intra-operative imaging such as the MRI probe 1 of FIG. 1. For example, while the small organ dedicated MRI probe 1 may be operated within a large shielded RF room, this will necessitate the use of special expensive shielded surgical equipment that is designed to create minimal RFI disturbances so as not to interfere with the operation of the MRI probe 1.

Brief Summary Text (22):
There is therefore provided, in accordance with a preferred embodiment of the present invention, electromagnetic apparatus for use in an MRI device. The probe includes a first permanent magnet assembly having a first surface and a second surface thereof. The probe also includes a second permanent magnet assembly having a

third surface and a fourth surface thereof. The second permanent magnet assembly opposes the first permanent magnet assembly such that the second surface and the third surface define an open region therebetween, for producing a predetermined volume of substantially uniform magnetic field extending in a first direction parallel to a first axis. The volume is disposed within the open region.

Brief Summary Text (23):
The probe also includes an energizable transmitting RF coil for producing an RF electromagnetic field within the volume, an energizable z-gradient coil for producing a magnetic field gradient extending within the open region in the first direction and parallel to the first axis, an energizable x-gradient coil for producing a magnetic field gradient extending within the open region in parallel to a second axis orthogonal to the first axis, and an energizable y-gradient coil for producing a magnetic field gradient extending within the open region in parallel to a third axis orthogonal to the first axis and the second axis. At least one of the x-gradient coil, y-gradient coil and z-gradient coil is positioned outside of the open region.

Brief Summary Text (24):
Furthermore, in accordance with another preferred embodiment of the present invention, the transmitting RF coil includes at least a first portion thereof positioned within the open region adjacent the second surface and at least a second portion thereof positioned within the open region adjacent the third surface. The first portion and the second portion of the transmitting RF coil are electrically connected in series.

Brief Summary Text (25):
Furthermore, in accordance with yet another preferred embodiment of the present invention, the transmitting RF coil further includes a third portion thereof including current return conductors positioned outside of the open region and adjacent the first surface, and at least a fourth portion thereof including current return conductors positioned outside of the open region and adjacent the fourth surface to increase the efficiency of the transmitting RF coil. The first portion, second portion, third portion and fourth portion of the transmitting RF coil are electrically connected in series.

Brief Summary Text (27):
Furthermore, in accordance with another preferred embodiment of the present invention, the shim coil includes a first shim coil portion positioned outside of the open region and opposed to the first surface of the first permanent magnet assembly, and a second shim coil portion positioned outside of the open region and opposed to the fourth surface of the second permanent magnet assembly.

Brief Summary Text (31):
Furthermore, in accordance with another preferred embodiment of the present invention, the first coil portion and the second coil portion of at least one of the x-gradient coil, y-gradient coil and z-gradient coil are substantially planar printed circuits, the first coil portion is assembled into a first multi-layer printed circuit assembly opposed to the first surface, and the second coil portion is assembled into a second multi-layer printed circuit assembly opposed to the fourth surface.

Brief Summary Text (34):
Furthermore, in accordance with another preferred embodiment of the present invention, the first permanent magnet assembly includes a first annular permanent magnet with a first and a second surface thereof. The first surface of the first annular permanent magnet is of a first magnetic polarity and the second surface of the first annular permanent magnet is of a second magnetic polarity. The first annular permanent magnet has an inside diameter. The first annular permanent magnet has at least a portion of the first surface of the first annular magnet lying in a first plane to provide a first magnetic field in the open region. The first magnetic field has a zero rate of change in a first direction at a first point in the open region. The first magnet assembly also includes at least a second annular permanent magnet with a first and a second surface thereof. The first surface of the second annular magnet is of the first magnetic polarity and the second surface of the second annular permanent magnet is of the second magnetic polarity. The second annular permanent magnet has an outside diameter which is smaller than the inside diameter of the first annular permanent magnet, with at least a portion of the first surface of the second annular magnet lying in a second plane spaced from the first

plane to provide a second magnetic field whereby the second magnetic field is superimposed upon the first magnetic field in the open region, having a zero rate of change in the first direction at a second point different from the first point. The second permanent magnet assembly includes a third annular permanent magnet with a first and a second surface thereof, the first surface of the third annular permanent magnet is of the second magnetic polarity and the second surface of the third annular permanent magnet is of the first magnetic polarity. The third annular permanent magnet has an inside diameter, the third annular permanent magnet has at least a portion of the first surface of the third annular magnet lying in a third plane to provide a third magnetic field, whereby the third magnetic field is superimposed on the first and second magnetic fields in the open region, having a zero rate of change in the first direction at a third point different from the first and second points. The second magnet assembly also includes at least a fourth annular permanent magnet having a first and a second surface thereof, the first surface of the fourth annular magnet is of the second magnetic polarity and the second surface of the fourth annular permanent magnet is of the first magnetic polarity. The fourth annular permanent magnet has an outside diameter which is smaller than the inside diameter of the third annular permanent magnet, with at least a portion of the first surface of the fourth annular permanent magnet lying in a fourth plane spaced from the third plane to provide a fourth magnetic field, whereby the fourth magnetic field is superimposed upon the first, second and third magnetic fields, in the open region, having a zero rate of change in the first direction at a fourth point different from the first, second and third points.

Brief Summary Text (42):
Furthermore, in accordance with another preferred embodiment of the present invention, the apparatus further including at least one receiving RF coil placeable adjacent to an organ or body part disposed within the open region.

Brief Summary Text (48):
Furthermore, in accordance with another preferred embodiment of the present intention, the x-gradient coil, the y-gradient coil and the z-gradient coil are planar printed circuit coil boards assembled within a single multi-layer printed circuit assembly positioned underneath the first permanent magnet assembly and the second permanent magnet assembly.

Brief Summary Text (49):
There is further provided, in accordance with a preferred embodiment of the present invention, electromagnetic apparatus for use in an MRI device. The apparatus includes a permanent magnet assembly having at least a first surface defining a first side of the permanent magnet assembly and a second surface defining a second side of the permanent magnet assembly opposed to the first side, for producing a predetermined volume of substantially uniform magnetic field extending in a first direction beyond the first surface. The apparatus further includes an energizable transmitting RF coil for producing an RF electromagnetic field within the volume. At least a portion of the RF coil is positioned adjacent the first surface of the permanent magnet assembly. The apparatus also includes an energizable z-gradient coil for producing a magnetic field gradient extending within the volume in the first direction parallel to a first axis. The apparatus also includes an energizable x-gradient coil for producing a magnetic field gradient extending within the volume parallel to a second axis orthogonal to the first axis. The apparatus also includes an energizable y-gradient coil for producing a magnetic field gradient extending within the volume parallel to a third axis orthogonal to the first axis and to the second axis. At least one of the x-gradient coil, y-gradient coil and z-gradient coil is positioned opposing the second surface of the permanent magnet assembly.

Brief Summary Text (52):
Furthermore, in accordance with another preferred embodiment of the present invention, the x-gradient coil, the y-gradient coil and the z-gradient coil are substantially planar printed circuits assembled within a substantially planar multi-layer printed circuit assembly. The multi-layer printed circuit assembly is disposed on the second side of the permanent magnet assembly facing the second surface.

Brief Summary Text (66):
There is also provided, in accordance with another preferred embodiment of the present invention, electromagnetic apparatus for use in an MRI device. The apparatus includes a permanent magnet assembly having a first surface and a second surface for producing a predetermined volume having a magnetic field varying substantially

linearly along a first axis. The volume extends in a first direction beyond the first surface along the first axis. The magnetic field is substantially uniform in any plane which is included within the predetermined volume and which is orthogonal to the first direction within the predetermined volume. The apparatus further includes an energizable transmitting RF coil for transmitting RF radiation. The RF coil has at least one portion thereof positioned opposing the first surface of the permanent magnet assembly. The apparatus also includes an energizable x-gradient coil for producing a magnetic field gradient along a second axis orthogonal to the first axis. The apparatus also includes an energizable y-gradient. coil for producing a magnetic field gradient along a third axis orthogonal to the first axis and to the second axis. At least one of the x-gradient coil and y-gradient coil is positioned opposing the second surface of the permanent magnet assembly.

Brief Summary Text (68):
There is also provided, in accordance with another preferred embodiment of the present invention, a method for constructing electromagnetic apparatus for use in an MRI device. The method includes the steps of providing a first permanent magnet assembly having a first surface and a second surface thereof, providing a second permanent magnet assembly having a third surface and a fourth surface thereof, positioning the second permanent magnet assembly opposite the first permanent magnet assembly such that the second surface and the third surface define an open region therebetween, for producing a predetermined volume of substantially uniform magnetic field extending in a first direction parallel to a first axis, the volume is disposed within the open region, providing an energizable transmitting RF coil for producing an RF electromagnetic field within the volume, providing an energizable z-gradient coil for producing a magnetic field gradient extending within the open region in the first direction and parallel to the first axis, providing an energizable x-gradient coil for producing a magnetic field gradient extending within the open region in parallel to a second axis orthogonal to the first axis, providing an energizable y-gradient coil for producing a magnetic field gradient extending within the open region in parallel to a third axis orthogonal to the first axis and the second axis, providing at least one receiving RF coil placeable adjacent to an organ or body part to be imaged for receiving RF signals from the organ or body part, and positioning at least one of the x-gradient coil, y-gradient coil and z-gradient coil outside of the open region for reducing the loading of the transmitting RF coil and the at least one receiving RF coil by the at least one of the x-gradient coil, y-gradient coil and z-gradient coil.

Brief Summary Text (69):
There is further provided, in accordance with another preferred embodiment of the present invention, a method for constructing electromagnetic apparatus for use in an MRI device. The method includes the steps of providing a permanent magnet assembly having at least a first surface defining a first side of the permanent magnet assembly and a second surface defining a second side of the permanent magnet assembly opposed to the first side, for producing a predetermined volume of substantially uniform magnetic field extending in a first direction beyond the first surface, providing an energizable transmitting RF coil for producing an RF electromagnetic field within the volume, positioning at least a portion of the transmitting RF coil adjacent the first surface of the permanent magnet assembly, providing at least one receiving RF coil placeable adjacent to an organ or body part to be imaged for receiving RF signals from the organ or body part, providing an energizable z-gradient coil for producing a magnetic field gradient extending within the volume in the first direction parallel to a first axis, providing an energizable x-gradient coil for producing a magnetic field gradient extending within the volume parallel to a second axis orthogonal to the first axis, providing an energizable y-gradient coil for producing a magnetic field gradient extending within the volume parallel to a third axis orthogonal to the first axis and to the second axis, and positioning at least one of the x-gradient coil, y-gradient coil and z-gradient coil opposite the second surface of the permanent magnet assembly for reducing the loading of the transmitting RF coil and the at least one receiving RF coil by the at least one of the x-gradient coil, y-gradient coil and z-gradient coil.

Brief Summary Text (70):
Finally, there is provided, in accordance with another preferred embodiment of the present invention, a method for constructing electromagnetic apparatus for use in an MRI device. The method includes the steps of providing a permanent magnet assembly having a first surface and a second surface for producing a predetermined volume having a magnetic field varying substantially linearly along a first axis, the volume extends in a first direction beyond the first surface along the first axis,

the magnetic field is substantially uniform in any plane included within the predetermined volume and orthogonal to the first direction within the predetermined volume, providing an energizable transmitting RF coil for transmitting RF radiation, positioning the transmitting RF coil such that at least one portion thereof opposes the first surface of the permanent magnet assembly, providing at least one receiving RF coil placeable adjacent to an organ or body part to be imaged for receiving RF signals from the organ or body part, providing an energizable x-gradient coil for producing a magnetic field gradient along a second axis orthogonal to the first axis, providing an energizable y-gradient coil for producing a magnetic field gradient along a third axis orthogonal to the first axis and to the second axis, and positioning at least one of the x-gradient coil and y-gradient coil opposite the second surface of the permanent magnet assembly for reducing the loading of the transmitting RF coil and the at least one receiving RF coil by the at least one of the x-gradient coil and y-gradient coil.

Drawing Description Text (4):
FIG. 1 is a schematic perspective view of a small organ dedicated MRI probe useful in brain surgery;

Drawing Description Text (6):
FIG. 3 is a schematic cross section illustrating part of a prior art MRI device using permanent magnets;

Drawing Description Text (7):
FIG. 4 is an isometric view illustrating part of an MRI probe using permanent magnets and having external gradient coils, in accordance with a preferred embodiment of the present invention;

Drawing Description Text (8):
FIGS. 5-7 are front views schematically illustrating printed circuit layout designs for an x-coil, y-coil and z-coil, respectively, useful in the MRI probe of FIG. 4;

Drawing Description Text (9):
FIG. 8 is a schematic isometric view illustrating part of an MRI probe having the z-gradient coil positioned in the volume between the two permanent magnet assemblies and the x and y gradient coils positioned outside of the volume between the two permanent magnet assemblies, in accordance with yet another preferred embodiments of the present invention;

Drawing Description Text (11):
FIG. 10 is a cross section illustrating part of an MRI probe having z-gradient coils positioned between two annular permanent magnets, in accordance with another preferred embodiment of the present invention;

Drawing Description Text (15):
FIG. 14 is a pictorial illustration of a small organ dedicated MRI probe used in conjunction with a local disposable RF cage, in accordance with a preferred embodiment of the present invention;

Drawing Description Text (16):
FIG. 15 is a schematic isometric view illustrating a transmitting RF coil providing linear polarization useful with the MRI probes of the present invention;

Drawing Description Text (17):
FIG. 16 is a schematic isometric view illustrating an MRI probe including the transmitting RF coil of FIG. 15 disposed therein, in accordance with a preferred embodiment of the present invention;

Drawing Description Text (18):
FIG. 17 is a schematic cross section of the MRI probe of FIG. 16 taken along the lines XVII--XVII;

Drawing Description Text (19):
FIG. 18 is a schematic cross section of an MRI probe having an internal Z-gradient coil and external X-gradient and Y-gradient coils, in accordance with another preferred embodiment of the present invention;

Drawing Description Text (20):
FIG. 19 is a schematic cross section of an MRI probe having an internal Z-gradient

coil and external X-gradient and Y-gradient coils, in accordance with yet another preferred embodiment of the present invention;

Drawing Description Text (21):
FIG. 20 is a schematic isometric view illustrating an RF coil combinable with the RF coil of FIG. 15 to form a circularly polarizing RF transmitting coil assembly for use with an MRI probe, in accordance with another embodiment of the present invention;

Drawing Description Text (23):
FIG. 22 is a schematic isometric view illustrating an MRI probe having external X, Y and Z-gradient coils, in accordance with still another preferred embodiment of the present invention;

Drawing Description Text (24):
FIG. 23 is a schematic diagram of an MRI probe having a single permanent magnet assembly, in accordance with yet another preferred embodiment of the present invention; and

Drawing Description Text (25):
FIG. 24 is a schematic diagram illustrating an MRI probe having a fixed magnetic field gradient, in accordance with another preferred embodiment of the present invention.

Detailed Description Text (2):
Reference is now made to FIG. 3 which is a schematic cross section illustrating part of a prior art MRI device 30 using permanent magnets. The MRI device 30 includes two permanent magnets 36 and 38. Each of the permanent magnets 36 and 38 is constructed from segments 34. The permanent magnets 36 and 34 are encased in a structure 32 made of a conducting metal such as soft iron and operating as a magnetic flux return circuit. The MRI device 30 further includes two multi-layer printed circuits 40 and 42 positioned in the volume between the two permanent magnets 36 and 38. The multi-layer printed circuits include the gradient coils. The MRI device 30 also includes RF coils (not shown) and the shim coils (not shown) of the MRI device 30. Each of the multi-layer printed circuits 40 and 42 is positioned in close proximity to the permanent magnets 36 and 38, respectively, such that enough room is left for positioning the organ 44 such as the knee or head of a patient between the multi-layer printed circuits 40 and 42.

Detailed Description Text (3):
Reference is now made to FIG. 4 which is an isometric view illustrating part of an MRI probe using permanent magnets and having external gradient coils, in accordance with a preferred embodiment of the present invention.

Detailed Description Text (4):
The part of the MRI probe illustrated in FIG. 4 includes the two annular permanent magnet assemblies 2 and 4 of FIG. 2 and two multi-layer printed circuit assemblies 52 and 54. The multi-layer printed circuit assemblies 52 and 54 each include x, y and z-gradient coils (not shown), and shim coils (not shown). The MRI probe of FIG. 4 also includes RF coils (not shown).

Detailed Description Text (5):
In contrast to the prior art Permanent magnet MRI device 30 of FIG. 3 in which the multi-layer printed circuits 40 and 42 including the gradient coils are positioned in the volume between the two permanent magnets 36 and 38, the multi-layer printed circuit assemblies 52 and 54 of FIG. 4 are positioned outside of the region 14 defined between the two permanent magnet assemblies 2 and 4.

Detailed Description Text (7):
Additionally, in prior art MRI systems having a large structure of an electrically conductive metal such as iron, which surrounds the magnet poles, the gradient coils cannot be placed outside of the magnet poles since the conductive metal will absorb most of the gradient field.

Detailed Description Text (14):
An advantage of placing the multi-layer printed circuit assemblies 52 and 54 outside the region 14, is the increase in available space in the region 14 which is used for accommodating the imaged or treated organ. For example, the multi-layer printed circuit assemblies 52 and 54 are sufficiently distanced from the magnet assemblies 2

and 4, respectively, to allow free space for the shoulders (not shown in FIG. 4) of a patient undergoing brain surgery.

Detailed Description Text (17):
Reference is now male to FIG. 8 which is a schematic isometric view illustrating part of an MRI probe having the z-gradient coil positioned in the volume between the two permanent magnet assemblies and having the x-gradient, y-gradient and the shim coils positioned outside of the volume between the two permanent magnet assemblies, in accordance with yet another preferred embodiment of the present invention.

Detailed Description Text (18):
The part of the MRI probe illustrated in FIG. 8 includes the two permanent magnet assemblies 2 and 4 of FIG. 2, a pair of multi-layer printed circuit assemblies 72 and 74 positioned outside of the region 14 and a pair of multi-layer printed circuit assemblies 76 and 78 positioned in the region 14 between the magnet assemblies 2 and 4. Each of the printed circuit boards 76 and 78 includes the z-gradient coils (not shown) as disclosed hereinabove. Each of the multi-layer printed circuit assemblies 72 and 74 includes the x and y gradient coils and the shim coils (not shown).

Detailed Description Text (22):
Reference is now made to FIG. 10 which is a cross section illustrating part of an MRI probe having z-gradient coils positioned between two annular permanent magnets, in accordance with another preferred embodiment of the present invention.

Detailed Description Text (23):
The magnet assembly 63 includes the annular permanent magnets 63a, 63b and 63c. In contrast to the MRI probe of FIG. 8 in which the z-gradient coils are substantially planar coils included in the multi-layer printed circuit assemblies 72 and 74, the z-gradient coil 65 of FIG. 10 is positioned between the annular permanent magnets 63a and 63b and extends in a direction parallel to the axis 12. The multi-layer printed circuit assembly 64 includes x-gradient and y-gradient coils (not shown) and the shim coil (not shown). The multi-layer printed circuit assembly 62 may also include the RF coils (not shown).

Detailed Description Text (40):
An advantage of the staggered double layer structure of the annular permanent magnet of FIG. 12 is the increased safety for the patient whose organ is imaged or treated in the MRI probe of the present invention.

Detailed Description Text (51):
U.S. Pat. No. 5,900,793 to Katznelson et al. disclosed hereinabove teaches a method of improving the homogeneity of the magnetic field between opposing annular permanent magnets used in an MRI probe. The method includes selecting segments from a batch of equi-angular segments so that the variations in a magnetic field strength of adjacent segments follow a cyclic curve having a regular period, and combining the segments to form an annular permanent magnet. A magnet assembly is formed by connecting two or more such annular permanent magnets by a low magnetic permeability material. Finally two such magnet assemblies are aligned such that for each pair of opposing annular permanent magnets, the cyclic curves representing the magnetic field variation are aligned in anti-phase. The method improves the homogeneity of the achievable permanent magnetic field.

Detailed Description Text (52):
In accordance with yet another preferred embodiment of the present invention, the method disclosed by Katznelson et al. in U.S. Pat. No. 5,900,793 can be similarly applied in constructing annular permanent magnets used in the MRI probes of the present invention.

Detailed Description Text (53):
Reference is now made to FIG. 14 which is a pictorial illustration of a small organ dedicated MRI probe used in conjunction with a local disposable RF cage, in accordance with a preferred embodiment of the present invention.

Detailed Description Text (54):
The small organ dedicated MRI probe 120 includes the magnet assemblies 2 and 4 of FIG. 1 which are attached to a surgical table 122 by an adjustable frame 123. The surgical table is made from a conductive material such as stainless steel. The probe 120 also include an RF cage 124. The RF cage 124 is made of a sheet of flexible conductive RF mesh having a size and shape suitable for covering the body of the

patient 6 and the magnet assemblies 2 and 4.

Detailed Description Text (55):
The RF cage 124 is electrically connected to the surgical table 122 for completing the shielding of the MRI probe 120. The RF cage 124 may be suitably grounded. The RF cage 124 also has an opening 126 therein. The opening 126 is used by the surgeons 130 and 132 for accessing the brain of the patient 6 during surgery. For example, a surgical instrument 128 can be inserted through the opening 126 into the brain of the patient 6. The size and shape of the opening 126 is designed to enable comfortable insertion and manipulation of surgical instruments therethrough, while still enabling effective shielding of the MRI probe creating a "self shielded" magnet.

Detailed Description Text (59):
Reference is now made to FIG. 15 which is a schematic isometric view illustrating a transmitting RF coil providing linear polarization useful with the MRI probes of the present invention.

Detailed Description Text (60):
The transmitting RF coil 140 is preferably made of a folded flat copper ribbon conductor but can be made of any other suitably shaped electrically conducting material capable of carrying the required electrical currents. The coil 140 includes four front conductor portions 142A, 142B, 144A and 144B. When the RF coil 140 is electrically energized, an electrical current flows therethrough in the direction indicated by the arrows. The four front conductors 142A, 142B, 144A and 144B effectively form an open Helmholz coil configuration suitable for generating a linearly polarized RF electromagnetic field.

Detailed Description Text (63):
Reference is now made to FIGS. 16 and 17. FIG. 16 is a schematic isometric view illustrating an MRI probe 150 including the transmitting RF coil 140 of FIG. 15 disposed therein, in accordance with a preferred embodiment of the present invention. FIG. 17 is a schematic cross section of the MRI probe 150 of FIG. 16 taken along the lines XVII--XVII.

Detailed Description Text (64):
The MRI probe 150 includes two permanent magnet assemblies 162 and 164. The permanent magnet assembly 162 includes a housing 182 and a set of three concentric annular permanent magnets 2A, 2B and 2C attached to the housing 182. The permanent magnet assembly 164 includes a housing 184 and a set of concentric annular permanent magnets 4A, 4B and 4C attached to the housing 184. The housings 182 and 184 are made of fiberglass or from any other suitable electrically non-conducting plastic material or the like. The details of the structure, construction and tuning of the annular permanent magnets included within the permanent magnet assemblies 162 and 164 are not the subject matter of the present invention and will therefore not be discussed in detail herein. The structure and design of such permanent magnet assemblies is disclosed in co-pending U.S. Pat. No. 5,900,793 to Katznelson et al.

Detailed Description Text (65):
The permanent magnet assembly 162 includes a first surface 182A and a second surface 182B. The permanent magnet assembly 164 includes a third surface 184A and a fourth surface 184B. The two permanent magnet assemblies 162 and 164 are attached to a frame 173 and oppose each other such that the second surface 182B and the third surface 184A define therebetween an open region of space 114. The permanent magnet assemblies 162 and 164 produce a region of substantially homogenous magnetic field 168 disposed within the region 114.

Detailed Description Text (66):
The probe 150 also includes multi-layer printed circuit assemblies 172 and 174. The printed circuit assemblies 172 and 174 each include planar printed circuits boards (not shown) comprising an X-gradient coil, a Y-gradient coil, a Z-gradient coil and shim coils as disclosed in detail hereinabove for the multi-layer printed circuit assemblies 52 and 54 the MRI probe of FIG. 4. The printed circuit assembly 172 is disposed outside of the region 114 and faces the first surface 182A of the permanent magnet assembly 162. The printed circuit assembly 174 is also disposed outside of the region 114 and faces the fourth surface 184B of the permanent magnet assembly 164.

Detailed Description Text (67):

The MRI probe 150 further includes a transmitting RF coil 140 for producing an RF electromagnetic field within the open region 114, and a receiving RF coil 175 positioned within the open region 114, adjacent to the organ or body part (not Shown) which is to be imaged, for receiving RF electromagnetic signals from the organ or body part.

Detailed Description Text (68):
It is noted that the receiving RF coil 175 can be any suitable type of receiving RF coil known in the art, such as a flexible RF coil (not shown) or other types of RF coils having suitable dimensions for positioning near the organ or body part which is being imaged. Furthermore, in accordance with another embodiment of the present invention the MRI probe 150 may also include a plurality of small receiving RF coils (not shown) which may be disposed at different positions adjacent the organ or body part (not shown) as is well known in the art.

Detailed Description Text (69):
The part of the transmitting RF coil 140 which includes the four front conductor portions 142A, 142B, 144A and 144B is disposed in the open region 114 between the permanent magnet assemblies 162 and 164.

Detailed Description Text (71):
Preferably, in accordance with a design for an open Helmholz coil the distance between the front conductor portions 144A and 144B and the distance between the front conductor portions 142A and 142B is designed such that .alpha.=60.degree., wherein .alpha. is the angle between the lines connecting the center point 169 of the imaging volume 168 with the centers of the front conductor portions 142A and 142B. The point 169 lies on the axis 12 and is the midpoint between the surfaces 182B and 184A. However, the angle .alpha. may also be different than 60.degree. depending, inter alia, on the particular design parameters of the transmitting RF coil.

Detailed Description Text (72):
The part of the transmitting RF coil 140 which includes the four current return conductor portions 152A, 152B, 154A and 154B is disposed outside of the open region 114. The current return conductor portions 152A and 152B are disposed between the surface 184B and the multi-layer printed circuit assembly 174, and the current return conductor portions 154A and 154B are disposed between the surface 182A, and the multi-layer printed circuit assembly 172.

Detailed Description Text (74):
An additional advantage of disposing the current return conductor portions 152A, 152B, 154A and 154B outside the region 114 is the increase in the space available within the open region 114 for positioning and manipulating an organ to be imaged or surgical instruments during medical interventional procedures.

Detailed Description Text (75):
It is noted that, the positioning of the multi-layer printed circuit assemblies 174 and 172 outside the region 114 and away from the front conductor portions 142A, 142B 144A and 144B, significantly reduces the loading of the transmitting RF coil 140 by the X, Y and Z coils (not shown) and the shim coils (not shown) which are disposed within the multi-layer printed circuit assemblies 174 and 172. The annular permanent magnets 2A, 2B, 2C, 4A, 4B and 4C have a lower electrical conductivity than the copper conductors of the X,Y,Z coils and the shim coils, because they are made of a material, such as an iron-neodimium-boron alloy, having electrical conductivity lower than copper and because of the construction of each of the annular permanent magnets 2A, 2B, 2C, 4A, 4B and 4C from a plurality segments which are electrically isolated from each other by an electrically non-conducting glue as disclosed in detail in co-pending U.S. Pat. No. 5,900,793 to Katznelson et al. Thus, the loading of the transmitting RF coil 140 and of the receiving RF coil 175 is significantly reduced by the placement of the multi-layer printed circuit assemblies 174 and 172 outside the region 114 and away from transmitting RF coil 140 and the receiving RF coil 175. The reduction in loading of the transmitting RF coil 140 enables achieving a desired transmitted signal quality without having to use expensive high-power RF transmitting Amplifiers. The reduction in loading of the receiving RF coil 175 enables achieving a significant improvement in the image quality obtained by the MRI probe 150.

Detailed Description Text (76):
It will be appreciated by those skilled in the art that, although the current return

conductor portions 152A, 152B and 154A, 154B are being positioned closer to the multi-layer printed circuit assemblies 174 and 172, respectively, by being disposed outside of the open region 114, thus, potentially increasing the loading of the RF coil 140 by the gradient coils and shim coils, the multi-layer printed circuit assemblies 174 and 172 can be sufficiently distanced from the current return conductor portions 152A, 152B and 154A, 154B, respectively, by moving the multi-layer printed circuit assemblies 174 and 172 along the axis 12 away from the point 169 to reduce the loading of the RF coil 140.

Detailed Description Text (77):
The design of the MRI probe 150 can be thus optimized to give a desired high image quality by reducing the loading of the RF coil 140 without having to excessively increase the distance of the multi-layer printed circuit assemblies 174 and 172 from the point 169 of the imaging volume 168 which will require the use of stronger and more expensive amplifiers to energize the gradient and shim coils.

Detailed Description Text (78):
It is noted that, in prior art large MRI devices such as whole body imaging MRI devices, the gradient coils and the transmitting RF coils are internally disposed in the region between the magnets. Typically, this region is large enough to allow designing a sufficient distance between the transmitting RF coil and the gradient coils, thus solving the problem of reducing the loading the RF coil by the gradient and/or shim coil.

Detailed Description Text (79):
In direct contrast, in the smaller and more compact MRI probes used in systems such as the interventional MRI/MRT systems of the present invention, the problem of loading of the transmitting RF coil is more difficult to solve because the region between the permanents magnets (such as the regions 14 and 114 of FIGS. 4 and 17, respectively) is small due to limitations on the allowable size of the permanent magnet assemblies. Thus, the use of external gradient and shim coils of the present invention which are placed outside the region between the magnet assemblies, has the advantage of making more space available between the permanent magnet assemblies as well as reducing the loading of the transmitting RF coil for improving the image quality attainable.

Detailed Description Text (80):
It is noted that, while in the MRI probe 150 of FIGS. 16 and 17 the X-gradient coil, Y-gradient coil, Z-gradient coil and shim coils are included within the multi-layer printed circuit assemblies 174 and 172 which are externally positioned outside the region 114, other preferred embodiments of the present invention are possible in which some of the gradient coils and/or the shim coils are internally positioned within the region between the permanent magnet assemblies 164 and 162.

Detailed Description Text (81):
Reference is now made to FIG. 18 which is a schematic cross section of an MRI probe 250 having an internal Z-gradient coil and external X-gradient and Y-gradient coils, in accordance with another preferred embodiment of the present invention. The MRI probe 250 includes two external multi-layer printed circuit assemblies 274 and 272 and two permanent magnet assemblies 262 and 264. The multi-layer printed circuit assemblies 274 and 272 are similar in construction to the multi-layer printed circuit assemblies 174 and 172 of FIG. 17, except that they do not include a Z-gradient coil. Thus, each of the multi-layer printed circuit assemblies 274 and 272 includes an X-gradient coil (not shown), a Y-gradient coil (not shown) and a shim coil (not shown).

Detailed Description Text (84):
Reference is now made to FIG. 19 which is a schematic cross section of an MRI probe 350 having an internal Z-gradient coil and external X-gradient and Y-gradient coils, in accordance with yet another preferred embodiment of the present invention.

Detailed Description Text (85):
The MRI probe 350 includes the multi-layer printed circuit assemblies 274 and 272 of FIG. 18 and permanent magnet assemblies 362 and 364. The permanent magnet assemblies 362 and 364 are identical to the permanent magnet assemblies 162 and 164 of FIG. 17 in all respects except that the permanent magnet assembly 362 also includes an extended Z-gradient coil 300 and that that the permanent magnet assembly 364 also includes an extended Z-gradient coil 302. The Z-gradient coil 300 is concentrically disposed between the annular permanent magnets 2A and 2B and the Z-gradient coil 302

is concentrically disposed between the annular permanent magnets 4A and 4B.

Detailed Description Text (91):
It is still further noted that, while the transmitting RF coil 140 of the MRI probe 150 is a linearly polarizing, other types of transmitting RF coils may be used.

Detailed Description Text (92):
Reference is now made to FIGS. 20 and 21. FIG. 20 is a schematic isometric view illustrating an RF coil 240 combinable with the RF coil 140 of FIG. 15 to form a circularly polarizing RF transmitting coil assembly for use with an MRI probe, in accordance with another embodiment of the present invention. FIG. 21 is a schematic isometric view illustrating a circularly polarizing RF transmitting coil assembly, assembled from the RF coil of FIG. 15 and the RF coil of FIG. 20.

Detailed Description Text (93):
The RF coil 240 of FIG. 20 is preferably made of a folded flat copper ribbon conductor but can be made of any other suitably shaped electrically conducting material capable of carrying the required electrical currents. The coil 240 includes four front conductor portions 242A, 242B, 244A and 244B. When the RF coil 240 is electrically energized, an electrical current flows therethrough in the direction indicated by the arrows. The four front conductors 242A, 242B, 244A and 244B effectively form an open Helmholz coil configuration.

Detailed Description Text (94):
The coil 240 also includes four current return conductor portions 252A, 252B, 254A and 254B. It is noted that, while the front conductor portions 142A, 142B, 144A and 144B and the current return conductor portions 152A, 152B, 154A and 154B of the transmitting RF coil 140 (FIG. 15) are aligned vertically, the front conductor portions 242A, 242B, 244A and 244B and the current return conductor portions 252A, 252B, 254A and 254B of the RF coil 240 are horizontally aligned.

Detailed Description Text (96):
The transmitting RF coil 240 of FIG. 20 may replace the transmitting RF coil 140 of the MRI probe 150 (FIG. 16). However, the transmitting RF coils 140 and 240 can also be combined to form the circularly polarizing transmitting RF coil 340 of FIG. 21.

Detailed Description Text (97):
In the transmitting RF coil 340, the front conductor portions 242A and 242B are aligned orthogonal to the front conductor portions 142A and 142B. The four front conductor portions 242A, 242B, 142A and 142B are disposed adjacent to the surface 184A of the permanent magnet assembly 164 (not shown) of the MRI probe. The front conductor portions 244A and 244B are aligned orthogonal to the front conductor portions 144A and 144B. The four front conductor portions 244A, 244B, 144A and 144B are disposed adjacent to the surface 182A of the permanent magnet assembly 162 (not shown) of the MRI probe.

Detailed Description Text (99):
The general design of circularly polarizing transmitting RF coils is known in the art as a quadrature-hybrid RF coil type. However, the inventors of the present invention have noted that by positioning some or all of the gradient coils and shim coils outside of the open region 114 the load on the transmitting RF coils can be significantly reduced and the RF coil efficiency is improved. Additionally, the positioning of the current return conductor portions 154A, 154B, 254A, 254B, 152A, 152B, 252A and 252B of the circularly polarizing transmitting RF coil 340 outside of the open region 114 additionally improves the RF coil efficiency by significantly increasing the distance of the current return conductor portions 154A, 154B, 254A, 254B, 152A, 152B, 252A and 252B from the open region 114.

Detailed Description Text (100):
It is noted that, while the transmitting RF coils 140 and 340 of FIGS. 15, 16 and 21 which are useful with the MRI probes of the present invention have the advantage that portions thereof such as the current return conductor portions are disposed outside the open region 114 to increase the space available therewithin, many other designs of linearly or circularly polarizing transmitting RF coils may be possibly used with MRI probes having external gradient coils disposed outside of the open region 114, which are within the scope and spirit of the present invention. For example, transmitting RF coils (not shown) in which all of the transmitting RF coil or coils are positioned within the open region 114 may also be used in embodiments of the present invention.

Detailed Description Text (101):
Reference is now made to FIG. 22 which is a schematic isometric view illustrating an MRI probe 450 having external X, Y and Z-gradient coils, in accordance with still another preferred embodiment of the present invention.

Detailed Description Text (102):
The MRI probe 450 include two opposed permanent magnet assemblies 462 and 464 defining an open region 414 therebetween. The permanent magnet assemblies 462 and 464 may be attached to or supported by one or more supporting structures such as a supporting frame (not shown for the sake of clarity of illustration) which is designed to enable access to the region 414 and to the head of the patient 6.

Detailed Description Text (103):
An organ or body part such as the head of a patient 6 may be positioned within the open region 414 for imaging. The permanent magnet assemblies 462 and 464 may be similar in design to the permanent magnet assemblies 162 and 164 of FIG. 16 but may also be any suitably designed pair of permanent magnet assemblies for providing a region of substantially homogenous magnetic field therebetween. The MRI probe 450 further includes a transmitting RF coil 440 which includes four portions 440a, 440B, 440C and 440D. the portions 440a, 440B, 440C and 440D of the RF coil 440 are printed circuit board assemblies which are suitably electrically connected (connections not shown), the copper conductors (not shown) included in the printed circuit board assemblies 440a, 440B, 440C and 440D are shaped in a similar way to the conductors of the RF coil 140 of FIG. 15. However, the transmitting RF coils 140or 340 of FIGS. 15 and 21, respectively, may also be used instead of the RF coil 440. The printed circuit board assemblies 440A and 440D may also include shim coils (not shown), however, the shim coils (not shown) may also be a pair of separate coils each disposed opposing one of the printed circuit board assemblies 440A and 440D at a distance therefrom.

Detailed Description Text (104):
The MRI probe 450 further includes a receiving RF coil 175 and a multi-layer printed circuit assembly 472.

Detailed Description Text (105):
The multi-layer printed circuit assembly 472 is disposed underneath the permanent magnet assemblies 462 and 464 and outside the open region 414. Thus, the region 414 may be relatively freely accessed.

Detailed Description Text (106):
The multi-layer printed circuit assembly 472 includes three printed circuits (not shown) including a X-gradient coil, a Y-gradient coil and a Z-gradient coil. It is noted that, since the relative positioning of the multi-layer printed circuit assembly 472 is different than the positioning of the multi-layer printed circuit assemblies 172 and 174 of FIG. 16, the design of the gradient coils is adapted to suit the different position of the coils relative to the direction of the main magnetic field. The positioning of the multi-layer printed circuit assembly 472 outside the open region 414 has the advantages of making more space available in the open region 414 and of reducing the loading of the RF transmitting coil 440 by increasing the spatial separation between the conducting gradient coil surfaces of the multi-layer printed circuit assembly 472 and the transmitting RF coil 440.

Detailed Description Text (107):
It is noted that while the MRI probes of the preferred embodiments disclosed hereinabove include a pair of opposing permanent magnet assemblies with an open region therebetween wherein an organ or body part is disposed in the open region between the pair of permanent magnet assemblies, other preferred embodiments of the present invention may be implemented using a single magnet assembly.

Detailed Description Text (108):
Reference is now made to FIG. 23 which is a schematic cross section of an MRI probe 500 having a single permanent magnet assembly, in accordance with yet another preferred embodiment of the present invention.

Detailed Description Text (109):
The MRI probe 500 includes a single permanent magnet assembly 562 having a first surface 582A and a second surface 582B opposing the first surface 582A. The permanent magnet assembly 562 may be constructed by using various different designs.

For example, the permanent magnet assembly 562 may be constructed from a plurality of concentric annular permanent magnets as disclosed in detail in co-pending U.S. Pat. No. 5,900,793 to Katznelson et al. However, the permanent magnet assembly 562 may also be implemented using other methods and designs adapted to provide a volume of substantially homogenous magnetic field 518 extending beyond the surface 582B of the permanent magnet assembly 562. The particular design parameters of the permanent magnet assembly may depend, inter alia, on the desired dimensions of the volume 518, the desired intensity of the magnetic field within the volume 518 and the distance between the volume 518 and the surface 582B.

Detailed Description Text (110):
The MRI probe 500 further includes a multi-layer printed circuit assembly 572. The multi-layer printed circuit assembly 572 is disposed opposing the surface 582A of the permanent magnet assembly 562 on the side of the permanent magnet assembly 562 which is opposite the side facing the volume 518. The multi-layer printed circuit assembly 572 includes three printed circuits (not shown) including a X-gradient coil, a Y-gradient coil and a Z-gradient coil. The multi-layer printed circuit assembly 572 may also include a shim coil (not shown) for active shimming of the main magnetic field.

Detailed Description Text (111):
The MRI probe 500 further includes a transmitting RF coil 540. The transmitting RF coil 540 is disposed between the surface 582B and the volume 518. The MRI probe 500 further includes a receiving RF coil 575 suitably connected to an RF amplifier 525 such as a low noise RF amplifier.

Detailed Description Text (113):
An advantage of the MRI probe 500 is that the gradient coils and shim coils which are included in the multi-layer printed circuit assembly 572 are disposed away from the region above the surface 582B and therefore do not restrict access of the imaged organ to the volume 518.

Detailed Description Text (114):
Another advantage of the configuration of the multi-layer printed circuit assembly 572 within the MRI probe 500 is that the gradient and shim coils (not shown) of the multi-layer printed circuit assembly 572 are positioned away from the transmitting RF coil 540 and the receiving RF coil 575 and therefore reduce the loading of the transmitting RF coil 540 and of the receiving RF coil 575 by the gradient coils (not shown) within the multi-layer printed circuit assembly 572, thereby improving image quality.

Detailed Description Text (119):
Reference is now made to FIG. 24 which is a schematic diagram illustrating an MRI probe having a fixed magnetic field gradient, in accordance with another preferred embodiment of the present invention.

Detailed Description Text (120):
The MRI probe 600 includes a single permanent magnet assembly 662 having a first surface 682A and a second surface 682B opposing the first surface 682A. The permanent magnet assembly 662 may be constructed by using various different designs. For example, the permanent magnet assembly 662 may be constructed from as a plurality of concentric annular permanent magnets as disclosed in detail in co-pending U.S. Pat. No. 5,900,793 to Katznelson et al., now U.S. Pat. 5,900,793 wherein the exact dimensions, shapes, magnetic field strength and relative positioning of the annular permanent magnets in the assembly are designed to obtain a fixed magnetic field gradient extending along the axis 612. This fixed Z-gradient varies substantially linearly within the predetermined volume 618 along the axis 612. The magnetic field is substantially uniform in any plane which is included within the volume 618 and is orthogonal to the axis 612 within the volume 618.

Detailed Description Text (122):
The MRI probe 600 further includes a multi-layer printed circuit assembly 672. The multi-layer printed circuit assembly 672 is disposed opposing the surface 682A of the permanent magnet assembly 662 on the side of the permanent magnet assembly 662 which is opposite the side facing the volume 618. In contrast to the multi-layer printed circuit assembly 572 of FIG. 23 which includes three gradient coils, the multi-layer printed circuit assembly 672 of FIG. 24 includes two printed circuits (not shown) including an X-gradient coil, and a Y-gradient coil. The multi-layer printed circuit assembly 672 may also include a shim coil (not shown) for active

shimming of the magnetic field.

Detailed Description Text (123):
The MRI probe 600 further includes a transmitting RF coil 640, The transmitting RF coil 640 is disposed between the surface 682B and the volume 618. The MRI probe 600 further includes a receiving RF coil 575 suitably connected to an RF amplifier 525 such as a low noise RF amplifier.

Detailed Description Text (126):
The advantages of the disclosed positioning of one or more of the gradient coils of the MRI probe 600 are similar to the advantages disclosed in detail for the MRI probe 500 hereinabove.

Detailed Description Text (127):
It is noted that, while the permanent magnet assemblies used within the MRI probes of FIGS. 4, 8-10, 14, 16-19, 22 and 23 are designed using concentric annular permanent magnets as disclosed in detail in co-pending U.S. Pat. No. 5,900,793 to Katznelson et al., many other types of magnet assemblies can be used which are within the scope of the present invention. For example, the annular permanent magnets used in the construction of the permanent magnet assemblies may be concentric polygonal annular shapes, or a plurality of elliptically shaped annuli having two common axes passing through the foci of the individual elliptical annuli.

Detailed Description Text (128):
Additionally, other configurations of permanent magnets may be used such as solid cube like or bar like permanent magnets or any other types of yoked or-non yoked magnets which are constructed to avoid the development of substantial eddy currents therewithin by the gradient coils. Such designs may use permanent magnetic materials having low electrical conductivity or may use magnetic and/or yoke structures which are segmented and are attached or glued by non-electrically conductive materials or glues. The development of eddy currents within yoke structures having high electrical conductivity may be reduced for enabling their use with the external gradient coils of the present invention by slotting the yoked structures with spiral or other types of slots to reduce possible current development. Thus, the various forms of the external gradient positioning of the present invention may be adapted for use with differently designed magnet assemblies configured singly, or as opposed pairs of magnetic assemblies having an open region therebetween.

Detailed Description Text (130):
It is still further noted that, in accordance with yet other preferred embodiments of the present invention; screening devices such as conducting metal mesh or grid may be inserted between various components of the MRI probes for improving RF screening. For example, in the MRI probe 150 of FIGS. 16 and 17, a suitable circular copper mesh piece (not shown) of a diameter similar to the diameter of the multi-layer printed circuit assembly 172 may be disposed between the surface 182A and the multi-layer printed circuit assembly 172, while another suitable circular copper mesh piece (not shown) of a diameter similar to the diameter of the multi-layer printed circuit assembly 174 may be disposed between the surface 184B and the multi-layer printed circuit assembly 174. Similarly, pieces of suitable copper mesh (not shown) may be used for screening the entire surface of the permanent magnet assembly 162 except the surface 182B thereof, and the entire surface of the permanent magnet assembly 164 except the surface 184A thereof.

Detailed Description Text (132):
It is further noted that, preferably, in all the embodiments of the MRI probes illustrated in FIGS. 4, 8-10, and 16-19, all the corresponding pairs of the gradient coils and shim coils of the MRI probe are electrically connected in series (the connections are not shown for the sake of clarity of illustration). For example, the Z-gradient amplifier (not shown) of the MRI probe 150 of FIG. 16 is electrically connected to the current input terminal (not shown) of the z-gradient coil (not shown) included within the multi-layer printed circuit assembly 172, the current output terminal (not shown) of the z-gradient coil of the multi-layer printed circuit assembly 172 is electrically connected to the current input terminal (not shown) of the z-gradient coil (not shown) included within the multi-layer printed circuit assembly 174, and the current output terminal (not shown) of the z-gradient coil of the multi-layer printed circuit assembly 174 is electrically connected to the Z-gradient amplifier, completing the circuit. Thus the z-gradient amplifier energizes both of the complementary z-gradient coils of the opposing multi-layer

printed circuit assemblies 172 and 174, simultaneously. A similar in-series electrical connection scheme is used for the pairs of complementary x-gradient coils (not shown) y-gradients (not shown) and Shim coils (not shown). However, other methods of connection of the complementary pairs of gradient and shim coils may also be used, such as the use of pairs of amplifiers (not shown), each of which activates one coil of the complementary pairs of coils.

CLAIMS:

1. Electromagnetic apparatus for use in an MRI device, the apparatus comprising:

a permanent magnet assembly having a first surface defining a first side of said permanent magnet assembly and a second surface defining a second side of said permanent magnet assembly opposed to said first side, for producing a predetermined volume of substantially uniform magnetic field extending in a first direction beyond said first surface;

an energizable transmitting RF coil for producing an RF electromagnetic field within said volume, at least a portion of said RF coil is positioned adjacent said first surface of said permanent magnet assembly;

an energizable z-gradient coil for producing a magnetic field gradient extending within said volume in said first direction parallel to a first axis;

an energizable x-gradient coil for producing a magnetic field gradient extending within said volume parallel to a second axis orthogonal to said first axis; and

an energizable y-gradient coil for producing a magnetic field gradient extending within said volume parallel to a third axis orthogonal to said first axis and to said second axis,

wherein at least one of said x-gradient coil, y-gradient coil and z-gradient coil is positioned opposing said second surface of said permanent magnet assembly.

5. The apparatus according to claim 1 wherein said x-gradient coil, said y-gradient coil and said z-gradient coil are substantially planar printed circuits assembled within a substantially planar multi-layer printed circuit assembly, said multi-layer printed circuit assembly is disposed on said second side of said permanent magnet assembly facing said second surface.

18. Electromagnetic apparatus for use in an MRI device, the apparatus comprising:

a permanent magnet assembly having a first surface and a second surface for producing a predetermined volume having a magnetic field varying substantially linearly along a first axis, said volume extending in a first direction beyond said first surface along said first axis, said magnetic field being substantially uniform in any plane included within said predetermined volume and orthogonal to said first direction within said predetermined volume;

an energizable transmitting RF coil for transmitting RF radiation, said RF coil having at least one portion thereof positioned opposing said first surface of said permanent magnet assembly;

an energizable x-gradient coil for producing a magnetic field gradient along a second axis orthogonal to said first axis; and

an energizable y-gradient coil for producing a magnetic field gradient along a third axis orthogonal to said first axis and to said second axis,

wherein at least one of said x-gradient coil and y-gradient coil is positioned opposing said second surface of said permanent magnet assembly.

20. A method for constructing an electromagnetic apparatus for use in an MRI device, the method comprising the steps of:

providing a permanent magnet assembly having at least a first surface defining a first side of said permanent magnet assembly and a second surface defining a second side of said permanent magnet assembly opposed to said first side, for producing a predetermined volume of substantially uniform magnetic field extending in a first

direction beyond said first surface;

providing an energizable transmitting RF coil for producing an RF electromagnetic field within said volume;

positioning at least a portion of said transmitting RF coil adjacent said first surface of said permanent magnet assembly;

providing at least one receiving RF coil placeable adjacent to an organ or body part to be imaged for receiving RF signals from said organ or body part;

providing an energizable z-gradient coil for producing a magnetic field gradient extending within said volume in said first direction parallel to a first axis;

providing an energizable x-gradient coil for producing a magnetic field gradient extending within said volume parallel to a second axis orthogonal to said first axis;

providing an energizable y-gradient coil for producing a magnetic field gradient extending within said volume parallel to a third axis orthogonal to said first axis and to said second axis; and

positioning at least one of said x-gradient coil, y-gradient coil and z-gradient coil opposite said second surface of said permanent magnet assembly for reducing the loading of said transmitting RF coil and said at least one receiving RF coil by said at least one of said x-gradient coil, y-gradient coil and z-gradient coil.

21. A method for constructing electromagnetic apparatus for use in an MRI device, the method comprising the steps of:

providing a permanent magnet assembly having a first surface and a second surface for producing a predetermined volume having a magnetic field varying substantially linearly along a first axis, said volume extending in a first direction beyond said first surface along said first axis, said magnetic field being substantially uniform in any plane included within said predetermined volume and orthogonal to said first direction within said predetermined volume;

providing an energizable transmitting RF coil for transmitting RF radiation;

positioning said transmitting RF coil such that at least one portion thereof opposes said first surface of said permanent magnet assembly;

providing at least one receiving RF coil placeable adjacent to an organ or body part to be imaged for receiving RF signals from said organ or body part;

providing an energizable x-gradient coil for producing a magnetic field gradient along a second axis orthogonal to said first axis;

providing an energizable y-gradient coil for producing a magnetic field gradient along a third axis orthogonal to said first axis and to said second axis; and

positioning at least one of said x-gradient coil and y-gradient coil opposite said second surface of said permanent magnet assembly for reducing the loading of said transmitting RF coil and said at least one receiving RF coil by said at least one of said x-gradient coil and y-gradient coil.

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L12: Entry 6 of 6 File: USPT Aug 30, 1994

DOCUMENT-IDENTIFIER: US 5343148 A
TITLE: Gradient coil system

Abstract Text (1):
In an NMR apparatus with a system of gradient coils for the production of a magnetic field gradient within a measurement volume (1) within a field coil, the transverse gradient coils are constructed unsymmetrically with respect to the z=0 plane dividing the measurement volume (1), however largely mirror symmetric to a x=0 or y=0 plane containing the z axis, and possess only two partial coils (20', 20''; 40', 40'') whose windings each exhibit the same winding direction. The axial gradient coils are cylindrically symmetric with respect to the z axis and totally unsymmetric with respect to the z=0 plane, and consist of at least two partial coils which are arranged on differing sides of z=0 plane, whereby the partial coils on one side largely exhibit the opposite winding direction than the coils on the other side, and whereby the number of coils with a particular winding direction is not equal to the number of coils with the opposite winding direction. In each case, it is thereby achieved that the measurement volume (1) is displaced to a, relative to the z axis, axial end of the apparatus so that access to the measurement volume (1) is substantially improved.

Brief Summary Text (2):
The invention concerns an apparatus for the production of nuclear magnetic resonance (NMR) with a field coil for the production of a homogeneous magnetic field B.sub.z in the direction of a z-axis and a system of gradient coils for the production of at least one approximately constant magnetic field gradient within a measurement volume in the homogeneous region of the magnetic field produced by the field coil, whereby the system of gradient coils exhibits at least a subsystem of transverse gradient coils for the production of magnetic field gradients G.sub.x or G.sub.y in a direction x or y perpendicular to the z-axis.

Brief Summary Text (4):
Such an NMR construction is by way of example known in the art from DE OS 31 33 873.

Brief Summary Text (5):
An important component of this type of NMR system, which is normally used for nuclear spin tomography, is a system of three gradient coils which, independent of another, are fed with currents of different strengths. These coils serve the purpose of overlapping constant field gradients of adjustable strengths upon the homogeneous magnetic field B.sub.0 of the main field coil, whereby the direction of one of these gradients (dB.sub.z /dz) is, as a rule, parallel to the direction of the homogeneous main field B.sub.0z, that is to say, to the axis of the main field magnet (z gradient=axial gradient) and the directions of the two other gradients (dB.sub.z /dx, dB.sub.z /dy) run orthogonal thereto and with respect to each other transverse to the direction of the main field (x and y gradients=transverse gradients). The spatial region in which the magnetic field of these gradient coils runs approximately linearly can be utilized for spatially resolved NMR methods (imaging, position selective spectroscopy) in so far as this region is not further limited through inhomogeneities of the main field. The gradient coils are, as a rule, attached to a cylindrical support pipe which surrounds the patient.

Brief Summary Text (6):
Due to the geometric configuration of conventional gradient coils, the support pipe has an axial extent to both sides of the center of the linear region which assumes a

value of 0.6 to 1.5 times the diameter of the support pipe. With the typical value of 0.7 m for the diameter of the support pipe, this is 0.42 m to 1.05 m. The patient is therefore surrounded by a relatively long narrow pipe. With sensitive patients, this can easily lead to conditions of claustrophobia. A further disadvantage of conventional gradient coils systems is due to the fact that is is not possible, in special body investigations as by way of example the examination of the head or the extremities, to utilize gradient coils in proximity to the object since the coils, due to the large axial separation from the linear region up to the ends of the support pipe, must surround the shoulders of the patient, that is to say must have a diameter of at least 0.5 m. Gradient coils in proximity to the object with small diameters had namely the advantage of a substantially reduced inductivity for a given gradient strength per unit current, through which correspondingly smaller gradient rise times were made possible. This is particularly advantageous for the executability of modern NMR examination methods (echoplanar methods etc.).

Brief Summary Text (10):
Towards this end it is therefore necessary that a patient, of which by way of example a tomogram of the inner skull is to be taken, be inserted head first over a long axial stretch along the z-axis into a long narrow pipe so that the head of the patient comes to rest in the measurement volume in the middle z=0 plane. In the event that the diameter d of the gradient coil, by way of example for the purpose of realizing high gradient strengths or small inductivity, is less than the shoulder width of the patient, the patient upon insertion into the magnet, would be stuck at the shoulders and the head would not be able to be brought at all into the axially distant measurement volume.

Brief Summary Text (11):
European laid open publication 0399789A2 discloses the use of a asymmetric transverse gradient coil in conjunction with fringe field imaging. In the fringe field imaging procedure, the axial gradient is generated by the inhomogeneous magnetic field at the edge of the magnet. Although these gradients can be extremely strong, their linearity is compromised. In conjunction with this system, symmetric axial gradient coils, to supplement the axial gradient of the fringe field, are disclosed. There is no mention of asymmetric shielding coils in conjunction with the system.

Brief Summary Text (12):
European laid open publication 0108421A2 discloses the use of an asymmetric axial gradient coil in conjunction with transverse coil systems for the purpose of displacing the center of the gradient region for imaging. Asymmetric transverse coils are not proposed nor are they relevant to this particular publication. The use of active shielding is discussed in Europan patent laid open publication 0433002A2 and a possible geometric configuration for asymmetric transverse gradient coils is suggested in the abstracts to the Society of Magnetic Resonance in Medicine Meeting, Aug. 1991, page 711.

Brief Summary Text (13):
In view of the above, it is therefore the purpose of the present invention to present an improved NMR apparatus that, on the one hand, compensates for patient claustrophobia while, on the other hand, allowing head examinations using an apparatus whose inner diameter clearance can be, for example, larger than the diameter of a human head, but smaller than the average shoulder width of a person.

Brief Summary Text (16):
In this fashion the separation z.sub.0 between the patient sided end of the gradient coil system and the center of the linear investigation region is substantially reduced. The geometry of the gradient coil system is, in the apparatus in accordance with the invention, so reformed that the region of linear magnetic field and gradient dependence useful for NMR investigations can already start on the patient end side of the gradient coil system. The detailed guiding of the winding sections on the patient side is thereby so carried out that a sufficiently larger region for NMR investigations results which exhibits a field dependence with sufficiently small deviations from ideal linear gradient field dependence.

Brief Summary Text (18):
The patient sided return section of a conventional gradient coil which, on the one hand, causes the large separation from the center of the linear region and, on the other hand, reduces in principle the size of the linear region suitable for NMR investigations, is displaced to the side turned away from the patient. The

unsymmetric perturbation in the field dependence thereby resulting in the vicinity of the zone provided for NMR investigations can, on the one hand, be kept small by displacing the return section of the windings at large distances from this region and, on the other hand, this perturbation can be largely compensated for by means of detailed guiding of the winding sections on the patient side.

Brief Summary Text (19):
Through the utilization of the transverse gradient coil in accordance with the invention it is possible, in a simple manner, to achieve the above mentioned purpose, namely the reduction of the distance z.sub.0 between the patient sided front side of the gradient coil system and the center of the measurement volume suitable for NMR investigations.

Brief Summary Text (20):
As a rule an apparatus for the excitation of nuclear magnetic resonances also contains a subsystem of axial gradient coils for the production of a magnetic field gradient G.sub.z in the direction of the z-axis. With the axial gradient coils, the axial extent to both sides of the center of the linear field region of the coil is less than that of transverse gradient coils. With coils of sufficient linearity, it assumes a value in general, of more than 0.44 times (.sqroot.3/4 times) the diameter of the coil. With the gradient coils which are suitable for whole body investigations of adults, and exhibit, in general, a diameter d of 0.7 m, this is about .+-. 0.3 m. This value is thereby significantly smaller than that of transverse gradient coils.

Brief Summary Text (22):
When utilizing the transverse gradient coils in accordance with the invention and conventional axial gradient coils, the above mentioned separation z.sub.0 between the patient sided front side of the gradient coil system and the center of the measurement volume suitable for NMR investigations is as a rule determined by the axial extent of the axial gradient coils.

Brief Summary Text (25):
When using the transverse coil system in accordance with the invention in conjunction with a conventional axial gradient coil system, the symmetric character of the axial conventional gradient system causes the axial extent of said system to project out beyond the asymmetrically arranged and foreshortened transverse gradient coils unless the separation of the conventional axial gradient coils is substantially less than usual. This, in turn, results in the conflicting requirements of maintaining open access through reduction of the total length of the transverse and axial gradient coils while maintaining good axial gradient coil linearity by keeping the partial axial gradient coils sufficiently far apart. Foreshortening the axial conventional symmetric gradient coils in order to prevent them from projecting out beyond the foreshortened asymmetric transverse gradient coils results in poor axial gradient linearity. It is therefore particularly advantageous to utilize the asymmetric axial gradient coils in conjunction with the asymmetric transverse gradient coils. This combination allows for maintenance of the requirement of open access while permitting sufficient linearity of both the axial and transverse gradient fields.

Brief Summary Text (26):
In embodiments of the invention, the transverse gradient coils can be asymmetrically arranged saddle coils with respect to the measurement volume, whereas in other embodiments are, with respect to the measurement volume, asymmetrically arranged streamline shaped coils which correspond to the conventional symmetrically arranged streamline shaped coils depicted in EP-0 320 285 with which an optimization of the coil inductivity n as well as a particularly high linearity of the corresponding field gradient can be achieved.

Brief Summary Text (27):
In a particularly advantageous embodiment of the NMR apparatus in accordance with the invention, a field coil is utilized for the production of the homogeneous magnetic field B.sub.0 which exhibits extremely small axial extent such as, by way of example, is described in EP-0 160 350. This so called "slice" field coil with very flat geometrical construction, has a homogeneity region up to the edge of the coil. When utilizing conventional gradient coils for tomographical reasons, this advantage goes largely unused, whereas with the modified asymmetric gradient coils in accordance with the invention the measurement region is taken advantage of up to the edge of the "slice" field coils which by way of example can be of great use in

mammography.

Brief Summary Text (28):
Since, when switching the electrical current through the above described asymmetric gradient coils, due to the lack of symmetry with respect to the z=0 plane dividing the measurement volume, correspondingly unsymmetric and perturbing eddy currents can establish themselves in the surrounding metal structure of the main field magnets, in a preferred embodiment of the invention, the gradient coils are actively shielded (see by way of example EP-A10 216 590). Towards this end, active shielding coils are arranged about the z axis on a cylinder which exhibits a larger radius than the cylinder containing the gradient coils, whereby the active shielding coils have largely the same symmetry properties as the gradient coils.

Brief Summary Text (29):
When utilizing the asymmetric transverse gradient coils in accordance with the invention, rapid switching of the gradient fields induces, through interaction with the main homogeneous magnetic field, Lorentz forces with these forces generating a torque on the transverse gradient coils. For a symmetric construction of the transverse gradient coils the two partial gradient coil halves, which are symmetric with respect to the central z=0 plane, generate torques in equal and opposite directions which, when the partial coil halves are rigidly conected to each other, cancel each other resulting in no net torque on the symmetric transverse gradient coil. However in the asymmetric case, there is no net cancellation of this torque and a large net torque on the asymmetric transverse gradient coil system thereby results. This torque is sufficiently large to present significant mechanical structure and mounting problems for the gradient coil system. However, when the asymmetric transverse gradient coils are used in conjunction with an asymmetric transverse coil active shield, the shielding currents will be in the opposite direction to those flowing in the transverse gradient coil and therefore the resulting torque on the shield will be opposite to that on the asymmetric transverse gradient coil. If the asymmetric transverse coil active shield and the asymmetric transverse gradient coil are rigidly attached to each other, the net torque on the combined system will be cancelled and the net combined mechanical system comprising the axial, the transverse asymmetric gradient coils, and their corresponding active shield will experience no net torque and therefore greatly improved mechanical stability.

Brief Summary Text (30):
In order to achieve a good shielding effect, it is necessary that the axial extent of the shielding coils be larger than the gradient coils themselves and, also in the direction of the patient side. Since, however, the active shielding coils have preferentially a diameter which is larger by approximately a factor of 1.15 to 1.4 than the gradient coils, in this fashion the good properties of the asymmetric gradient coil system in accordance with the invention with regard to the desired reduction of claustrophobia and the improved utilization of the measurement volume are not compromised.

Brief Summary Text (31):
In order to achieve an optimal shielding effect, in a preferred embodiment, the shielding coils are provided with an axial length on both sides of the z=0 plane which is larger than that of the gradient coils.

Brief Summary Text (32):
In particular with the utilization of a very flat "slice" field coil the windings of the shielding coils on the patient sided end of the apparatus would thereby, however, project out beyond the axial end of the gradient coil and thereby negate the advantage of the extremely small axial extension.

Brief Summary Text (33):
Preferentially for this case, in one embodiment, the shielding coil windings can be guided at the position (z=-.vertline.Z.sub.0 .vertline.) given by the axial end of the gradient coil in the z=-.vertline.Z.sub.0 .vertline. plane and be radially distributed. The distribution of the windings of the shielding coils in the z=-.vertline.z.sub.0 .vertline. plane as well as in the region z>-.vertline.z.sub.0 .vertline. over the cylinder surface containing the other parts of the shielding coils with radius R.sub.s is to be configured in such a manner that the magnetic field produced thereby through the gradient coils is minimized in the partial region z>-.vertline.z.sub.0 .vertline. at radius r>R.sub.s.

Brief Summary Text (34):
In this fashion the patient sided end of the shielding coil which would otherwise project in an axial direction onto the patient is, to a certain extent, radially "bent-off" in the outer direction and no longer impedes the axial access to the measurement volume.

Drawing Description Text (3):
FIG. 1a a schematic representation of the asymmetric saddle gradient coil in accordance with the invention with indicated spherically shaped measurement volume;

Drawing Description Text (4):
FIG. 1b conventional saddle shaped transverse gradient coils according to prior art;

Drawing Description Text (5):
FIG. 2a a flat winding of the double saddle transverse gradient coils in accordance with the invention with indicated position of the measurement volume and arrows in the direction of the current flow;

Drawing Description Text (6):
FIG. 2b a winding of a conventional symmetrically arranged double saddle coil according to prior art;

Drawing Description Text (7):
FIG. 3a a vertical cut through the NMR apparatus in the x=0 or y=0 plane with symmetric gradient coil system in accordance with the invention;

Drawing Description Text (10):
above: a schematic cross section through an NMR apparatus in accordance with the invention with asymmetric axial gradient coils,

Drawing Description Text (13):
above: a vertical cross section through a conventional NMR apparatus

Drawing Description Text (15):
FIG. 5 schematic of a vertical cross section through an NMR apparatus in accordance with the invention with asymmetric gradient coils and patient sided "bent-off" shielding coils.

Detailed Description Text (2):
The schematically represented configuration of FIG. 1b of conventional simple saddle shaped transverse gradient coils exhibits a mirror symmetry of the four partial coils 11', 11'', 11''' and 11'''' with respect to the symmetry planes z=0 and y=0. By means of current flowing in the direction of the arrow, a field gradient, which runs approximately linearly in the y direction, is produced in a measurement volume 1. The measurement volume 1 is approximately defined by a sphere whose center is at the point of intersection of the three symmetry planes Z=0, x=0 and y=0. For the production of the linear transverse field gradient in the measurement volume 1, only the "useful saddle portions" 12', 12'', 12''' and 12'''' contribute, whereas the remaining parts of the partial coils 11', 11'', 11''' and 11'''', in particular those of the "return current saddle parts" 13', 13'', 13''' and 13'''' which are turned away from the measurement volume serve solely to return the current through the partial coils. FIG. 1b clearly shows that the access to the measurement volume 1 in the direction of the z axis is in particular hindered by the return current saddle parts 13', 13'', 13''' and 13'''' which project away in the axial direction from the measurement volume.

Detailed Description Text (3):
In contrast thereto, with the asymmetric configuration of transverse gradient coils 8 in accordance with the invention shown in FIG. 1a, the return current saddle parts 23', 23'' of the associated useful saddle parts 22', 22'' of the partial coils 21', 21'' are folded as seen by the viewers to the right side of the measurement volume 1. Compared to the conventional configuration, the additional partial coils 21''', 21'''', which are likewise arranged on the right side of the z=0 plane remain unchanged. For geometrical reasons the saddle portions of the modified partial coils 21', 21'' surround these unchanged partial coils 21''', 21''''. Thereby, the partial coils 21', 21''' together form a partial coil 20' and the partial coils 21'', 21'''' together a partial coil 20''. In consequence of this the total configuration of the asymmetric transverse gradient coils 8 according to the invention is constructed

from solely two partial coils 20', 20'' which lie symmetrically across from another with respect to the y=0 plane. The fact that the useful saddle parts 21', 21'' produce a somewhat different field distribution than that of the somewhat smaller useful saddle portions 22''', 22'''' can be taken into account through differing winding numbers of the corresponding partial coils and through an adapting of the exact axial positions of the saddle portions 21', 21''. In this manner it is possible in measurement volume 1 to produce an approximately linear field gradient dependence. FIG. 1a clearly shows that the axial access to the measurement volume 1 in the z direction at the viewer's left side of the z=0 plane of the configuration is distinctly improved through the "folding away" of the return current saddle portions 23', 23''.

Detailed Description Text (4):
The schematic winding shown in FIG. 2b of a system of four conventional double saddle coils which produce a gradient field of higher linearity than that of the simple saddle coils shown in FIG. 1b, is, for its part, comprised of four partial coils 30', 30'', 30''' and 30''''. The measurement volume 1 is indicated twice in the view. Each of the double saddle coils is composed of a nesting of two partial coils 31', 31'' whose useful saddle portions 32', 32'' are held at an axial separation with respect to each other in the direction of the z axis and contribute, when current is flowing through the coils, to the construction of the transverse magnetic field gradient, whereas the corresponding return current saddle parts 33', 33'' are guided parallel and in proximity to another on the side axially removed from the measurement volume 1.

Detailed Description Text (5):
In contrast, the asymmetric subsystem of transverse gradient coils 8 in accordance with the invention shown in FIG. 2a is comprised of only two oppositely positioned partial coils 40', 40'' in which the useful saddle portions 42', 42'', 42''', 42''' associated with the return current saddle portions 43', 43'', 43''', 43'''' are folded in the direction of the side of the z=0 plane which is turned away from the useful saddle portions. In this fashion at the observer's left side of the asymmetrical double saddle coils 8 shown in FIG. 2a, a significantly simplified axial access to the measurement volume 1 in the z direction is again given since the separation z.sub.0 between the patient sided end of the gradient coil system and the center of the linear investigational region is substantially reduced compared to the conventional system.

Detailed Description Text (6):
The same effect also manifests itself when comparing FIG. 3b, where a conventional NMR system with symmetric gradient coils is schematically represented in transverse cross section, to the system in accordance with the invention of FIG. 3a.

Detailed Description Text (7):
Indicated here are the field coil 2, active shielding coils 3 and 6 each, and in FIG. 3b the symmetric axial gradient coils 4 or in FIG. 3a the modified asymmetric axial gradient coil 5 in accordance with the invention. Here also, for the case of the asymmetric gradient coils, the distance z.sub.0 is substantially reduced.

Detailed Description Text (8):
FIG. 4b again shows a schematic vertical section through a conventional NMR apparatus with field coil 2, active shield coils 3 and symmetrically arranged axial gradient coils 4 which enclose a measurement volume 1. Thereby, the anti-symmetric dependence with respect to the z=0 plane of the magnetic field gradients in the direction of the z axis thereby produced are also schematically shown. FIG. 4a on the other hand shows the asymmetric axial gradient coils 5 in accordance with the invention with the associated likewise asymmetrically arranged active shielding coil 6. The corresponding field dependence B.sub.z (z) is no longer anti-symmetric with the respect to the z=0 plane, rather is completely unsymmetric. Through corresponding configuration of the windings of the axial gradient coils 5 it is possible to achieve a sufficiently larger region for NMR investigations with only small deviations from ideal linear gradient dependence.

Detailed Description Text (9):
Since the intrinsic property of conventional anti-symmetric axial gradient coils 4 of a vanishing field value of B.sub.z= 0 in the plane z=0 is not a requirement for carrying out position selective NMR methods, the properties of the unsymmetric axial gradient coils 5 in accordance with the invention, with regard to a reduction of the distance z.sub.0 and/or an improvement in the linearity, lend themselves to further

improvement through the realization of a gradient field dependence with the following properties:

Detailed Description Text (10):
In the region of the measurement volume 1 provided for NMR investigations a field dependence of the form

Detailed Description Text (11):
is approximately realized. The coordinate z=0 thereby lies approximately in the center of the region provided for NMR investigations. Furthermore B.sub.0 and B.sub.1 /z.sub.1 both have the same sign and B.sub.1 approximately equal to B.sub.0. It is thereby possible to approximately realize the net field dependence given in the larger portion of FIG. 4a.

Detailed Description Text (12):
A further improvement of the apparatus in accordance with the invention is represented in FIG. 5 where the active shielding coils 6 which are matched to the symmetry of the totally unsymmetric axial gradient coils 5 do not, as in conventional coils, project in the z direction beyond the axial gradient coils in the region in proximity to the patient, rather are, in a certain sense, "bent-off" through radial distribution in a plane. Thereby a further improvement of the axial access to the measurement volume 1 or a reduction in the separation z.sub.0 results.

Detailed Description Text (13):
In the configuration of FIG. 5 the asymmetric transverse coils 8 are located, by way of example, within the axial asymmetric gradient coils 5. The mountings of the transverse asymmetric gradient coils 8, the axial asymmetric gradient coils 5, and the active shielding coils 6 are so arranged that all three sets of coils are rigidly connected to each other by means of a rigid connector means 9. The gradient coils could be epoxied together into one subsystem and the asymmetric active shielding coils could be epoxied together into a separate subsystem and rigidly attached to each other or the entire system of gradient plus shielding coils could be epoxied together as one solid unit. The rigid connector means 9 assure that the torques generated on the asymmetric transverse gradient coils 8 are cancelled at least partially through the oppositely directed torques on the corresponding active shielding coils 6.

CLAIMS:

1. An apparatus for the production of nuclear magnetic resonance (NMR) comprising:

(a) a field coil for the production of a homogeneous magnetic field B.sub.z in the direction of a z-axis, said homogeneous field having a measurement volume whose center lies on the z axis;

(b) a transverse gradient coil for the production of a magnetic field gradient in a direction perpendicular to the z axis, said transverse gradient coil being constructed to be unsymmetric with respect to a z=0 plane perpendicular to the z axis through the center of the measurement volume; and

(c) an axial gradient coil for the production of a magnetic field gradient G.sub.z in the direction of the z axis, said axial gradient coil being arranged cylindrically symmetric with respect to the z axis and unsymmetric with respect to the z=0 plane.

2. An apparatus according to claim 1, wherein said transverse gradient coil is, relative to the measurement volume, an asymmetrically arranged saddle coil.

5. An apparatus according to claim 1, further comprising:

(d) active shielding coils, said coils being arranged on a cylindrical surface about the z axis at a larger radius than said transverse and axial gradient coils, with said shielding coils having largely the same symmetry properties as said transverse and axial gradient coils.

6. An apparatus according to claim 5, wherein said shielding coils exhibit a diameter which is larger than that of said gradient coils by a factor of 1.15 to 1.4.

7. An apparatus according to claim 5, wherein said shielding coils are provided to have a larger axial length than said gradient coils on both sides of the z=0 plane.

8. An apparatus according to claim 5, wherein the axial extent of said shielding coils on the side of the z=0 plane lying away from a point on the z axis at the middle of said field coil is arranged to be at the axial end of said gradient coils, and the windings of said shielding coils are radially distributed at this axial end in a plane.

10. An apparatus according to claim 9, wherein said transverse gradient coil is, relative to the measurement volume, an asymmetrically arranged saddle coil.

13. An apparatus according to claim 9, further comprising active shielding coils, and coils being arranged on a cylindrical surface about the z axis at a larger radius than said transverse and axial gradient coils, said shielding coils having largely the same symmetry properties as said gradient coils.

14. An apparatus according to claim 13, wherein said shielding coils exhibit a diameter which is larger than that of said gradient coils by a factor of 1.15 to 1.4.

15. An apparatus according to claim 13, wherein said shielding coils are provided to have a larger axial length than said gradient coils on both sides of the z=0 plane.

16. An apparatus according to claim 13, wherein the axial extent of said shielding coils on the side of the z=0 plane lying away from a point on the z axis at the middle of said field coil is arranged to be at the axial end of said gradient coils, and the windings of said shielding coils are radially distributed at this axial end in a plane.

DOCUMENT-IDENTIFIER: US 5497089 A
TITLE: Wide aperture gradient set

Abstract Text (1):
An insertable coil (40) is inserted in a bore (12) of a magnetic resonance imaging apparatus. Primary field magnets (10) create a temporally constant magnetic field longitudinally through the insertable coil. A computer control (58) controls a radio frequency coil (44) and a gradient coil (42) to create magnetic resonance imaging sequences and process received magnetic resonance signals into image representations. The insertable gradient coil includes a central cylindrical portion (60) having a first circumference. A second portion (62) disposed toward a patient receiving end of the insertable coil has a second circumference which is larger than the first circumference. In this manner, the first, smaller circumferential portion is adapted to receive the patient's head and the larger circumferential portion is adapted to accommodate the patient's shoulders. For symmetry which eliminates magnetic field induced torques, a service end (68) matches and is symmetric to the patient end (62). The z-gradient coil windings (FIG. 3), and x and y-gradient coil windings (FIGS. 4 or 5) are mounted such that a portion of the windings are on the first, smaller circumferential portion and a portion of the windings are on the larger patient and service end portions.

Brief Summary Text (2):
The present invention relates to the magnetic resonance imaging art. It finds particular application in conjunction with insertable gradient coils for high speed imaging techniques and will be described with particular reference thereto.

Brief Summary Text (3):
Magnetic resonance imagers commonly include a large diameter, whole body gradient coil which surrounds a patient receiving bore. Main field magnets, either superconducting or resistive, and radio frequency transmission/reception coils also surround the bore. Although the whole body gradient coils produce excellent linear magnetic field gradients, they have several drawbacks. With large diameter gradient coils, the slew rate is sufficiently slow that it is a limiting factor on the rate at which gradient magnetic fields can be induced and changed. Large diameter whole body gradient coils are too slow for some of the highest speed magnetic resonance imaging techniques. The energy stored in gradient coils is generally proportional to greater than the fifth power of the radius. Hence, large diameter, whole body coils require large amounts of energy. Further, superconducting main magnets have cold shields disposed around the bore. The larger the diameter of the gradient coil, the closer it is to the cold shields and hence the more apt it is to produce eddy currents. More shielding is needed to prevent the whole body gradient coils from inducing eddy currents in the cold shields than would be necessary for smaller diameter coils.

Brief Summary Text (5):
As a general rule, the longer the cylindrical head coil, the larger the region over which the gradient is linear and the more linear the region is. However, the patient's shoulders are a limiting factor on the length of a symmetric gradient coil. The shoulders limit the isocenter to about 20 cm at the patient end. Thus, symmetric head coils have heretofore been limited to about 40 cm in length.

Brief Summary Text (7):
Although conventional head gradient coils include a Maxwell pair for the z-axis or Golay saddle coils for the x or y-axes on the surface of a cylinder, others have proposed coils in which all windings do not lie on the cylinder surface. "Compact

Magnet and Gradient System For Breast Imaging", S. Pissanetzky, et al., SMRM 12th Annual Meeting, p. 1304 (1993) illustrates a compact asymmetric cylinder coil bent up radially at a 90.degree. angle at the field producing end of the coil. The coil is designed for breast imaging with the coil pressed up against the chest. "High-Order, Multi-Dimensional Design of Distributed Surface Gradient Coil", Oh et al., SMRM 12th Annual Meeting, p. 310 (1993) attempts to optimize a gradient surface coil using current flows in three dimensions. One of the problems with the Oh surface gradient coil is that it was difficult to control linearity. Further, the coil was difficult to manufacture due to its complicated shape and high current densities.

Brief Summary Text (12):
In accordance with another more limited aspect of the present invention, the gradient coils include a plurality of thumbprint windings. The thumbprint windings have the first, smaller radius adjacent the isocenter and are flared to the second, larger radius adjacent at least the patient end of the coil.

Drawing Description Text (3):
FIG. 1 is a diagrammatic illustration of a magnetic resonance imaging system including an insertable coil in accordance with the present invention;

Drawing Description Text (6):
FIG. 4 is a diagrammatic illustration of one quadrant of an x or y-gradient fingerprint type winding for the gradient coil of FIG. 3 laid out flat;

Detailed Description Text (3):
A whole body gradient coil assembly 30 includes x, y, and z-coils mounted along the bore 12. Preferably, the gradient coil assembly is a self-shielded gradient coil assembly that includes primary x, y, and z-coil assemblies potted in a dielectric former 32 and a secondary gradient coil assembly 34 that is supported on a bore defining cylinder of the vacuum dewar 22. A whole body RF coil 36 is mounted inside the gradient coil assembly 30. A whole body RF shield 38, e.g. copper mesh, is mounted between RF coil 36 and the gradient coil assembly 34.

Detailed Description Text (4):
An insertable gradient coil 40 is removably mounted in the center of the bore 12. The insertable coil assembly includes an insertable gradient coil assembly 42 supported by a dielectric former. An insertable RF coil 44 is mounted inside the dielectric former. An RF shield 46 is mounted between the insertable RF and gradient coils.

Detailed Description Text (5):
An operator interface and control station 50 includes a human-readable display such as a video monitor 52 and an operator input means including a keyboard 54 and a mouse 56. A computer control and reconstruction module 58 includes computer hardware and software for controlling the radio frequency coils 36 and 44 and the gradient coils 30 and 42 to implement any of a multiplicity of conventional magnetic resonance imaging sequences, including echo-planar imaging sequences. Echo-planar imaging sequences are characterized by short repetition rates and low flip angles. The processor 58 also includes a digital transmitter for providing RF excitation and resonance manipulation signals to the RF coil and a digital receiver for receiving and demodulating magnetic resonance signals. An array processor and associated software reconstruct the received magnetic resonance signals into an image representation which is stored in computer memory or on disk. A video processor selectively extracts portions of the stored reconstructed image representation and formats the data for display by the video monitor 52.

Detailed Description Text (6):
With reference to FIG. 2, the active gradient coil windings of the insertable gradient coil assembly 42 in the preferred embodiment are confined to a first cylindrical surface region 60 and an open patient end conical surface 62. The cylindrical surface 60 has an isocenter 64 mid-way, a distance z.sub.cp from either edge of the cylindrical surface. The cylindrical surface has an interior dimension sized to receive the human head, preferably with a radius .rho..sub.c equal to about 15 cm. The RF screen and RF head coil reside inside of this diameter, between the subject's head and the gradient. The conical section 62 and a matching service end conical portion 68 flare from the radius .rho..sub.c to a larger radius .rho..sub.s.

Detailed Description Text (19):
Various alternate embodiments are also contemplated. With reference to FIG. 6, the coil may again have a central cylindrical portion 60 sized to fit the subject's head or other anatomical portion. A flared region 62 connects the central cylindrical region with outer cylindrical regions 66 of larger diameter. In a head coil embodiment, the larger cylindrical region 66 is of sufficient dimension to receive the patient's shoulders therein. For symmetry purposes, matching flared regions and cylindrical regions are preferably provided on both the patient and the service end 68 of the head coil. As another alternative, the central cylinder 60 can be elliptical to follow the generally elliptical cross-section of the human head and the outer larger cylindrical portion 66 is elliptical in a direction to match the aspect ratio of the human shoulders.

Other Reference Publication (3):
"NMR Imaging in Biomedicine, Supplement 2," Mansfield, et al., pp. 268-269 (1982).

CLAIMS:

1. In an insertable gradient coil for a magnetic resonance imaging apparatus, which insertable gradient coil includes a central cylindrical portion having an interior bore of a first cross-section dimensioned to receive a head of the subject to be imaged, the central cylindrical portion carrying gradient coil windings for creating linear magnetic field gradients in the interior bore through the subject's head, THE IMPROVEMENT COMPRISING:

the insertable gradient coil further including a first end portion of a second cross-section dimensioned to receive shoulders of the subject to be imaged, the second cross-section being larger than the first cross-section, the first end portion being connected with a first end of the central cylindrical portion adjacent a patient receiving end of the insertable gradient coil, the gradient coil windings extending in part along the first end portion such that the gradient coil windings lie in part on the central cylindrical portion and in part off the central cylindrical portion on the first end portion.

3. A gradient magnetic field coil assembly for a magnetic resonance imaging apparatus, the gradient field coil assembly comprising:

a central cylindrical region of a first circumference;

a first end region adjacent at least one end of the central region having a second circumference, the second circumference being larger than the first circumference;

a gradient coil for generating magnetic field gradients within the smaller circumference portion, the gradient coil having windings on both the smaller circumference central region and the larger circumference first end region.

4. The gradient coil assembly as set forth in claim 3 further including a second end region disposed adjacent a second end of the central cylindrical region, the second end region having a circumference that is larger than said first circumference.

5. The gradient coil assembly as set forth in claim 3 wherein the gradient windings include an axial gradient winding extending circumferentially around the central cylindrical region with a plurality of turns and around the first end region with a further plurality of turns for generating gradient magnetic fields axially along the coil.

6. A gradient magnetic field coil assembly for a magnetic resonance imaging apparatus, the gradient field coil assembly comprising:

a central region of a first cross-section;

a first end region adjacent a first end of the central region, the first end region having a second circumference that is larger than the first circumference;

a set of four gradient windings for generating a magnetic field gradient component within the central region across a central axis thereof, at least two of the gradient windings being bunched coils which include arc segments extending along the central region and return paths extending along the first end region.

7. A gradient magnetic field coil assembly for a magnetic resonance imaging apparatus, the gradient field coil assembly comprising:

a first region extending around a longitudinal axis;

a second region extending around the longitudinal axis and being disposed adjacent at least one end of the first region, the second region being disposed further from the longitudinal axis than the first region;

a set of gradient coil windings for generating a magnetic field gradient across the longitudinal axis, the winding set including at least first and second distributed coil windings each with its windings extending in generally a thumbprint pattern, the first and second distributed coil winding each having their thumbprint pattern disposed in part on the first region and in part on the second region.

8. A gradient magnetic field coil assembly for a magnetic resonance imaging apparatus, the gradient field coil assembly comprising:

a central cylindrical region of a first, smaller circumference;

an end region adjacent at least one end of the central region having a second, larger circumference, the second circumference being larger than the first circumference;

the smaller circumference central cylindrical region and the larger circumferential end region both being circularly symmetric;

a gradient coil for generating magnetic field gradients within the smaller circumferential central cylindrical portion, the gradient field coil having windings on both the smaller circumferential central cylindrical region and the larger circumferential end region.

9. The gradient coil assembly as set forth in claim 3 wherein the larger circumferential end region includes a flared portion which angles outward from the smaller diameter central region along an open, generally conical segment.

10. The gradient coil assembly as set forth in claim 3 wherein the larger circumferential end region includes a cylindrical portion.

11. A magnetic resonance imaging system comprising:

a primary magnetic field assembly for generating a temporally constant magnetic field through a central bore thereof;

a whole body gradient coil assembly disposed around the central bore;

a radio frequency coil assembly disposed around the central bore;

a head coil removably mounted in the central bore, the head coil including:

a plurality of gradient coil windings which extend around a first cylindrical portion of a first circumference and around at least a second portion of a second circumference, which second circumference is larger than the first circumference, the second circumference portion being mounted to a patient receiving end of the head gradient coil, and

an RF coil winding disposed within at least the first cylindrical portion;

a magnetic resonance excitation and reconstruction means for controlling the radio frequency and insertable gradient coils for inducing magnetic resonance within the first cylindrical portion and for receiving magnetic resonance signals therefrom and for reconstructing the received magnetic resonance signals into an image representation.

12. A magnetic resonance imaging system comprising:

a primary magnetic field assembly for generating a temporally constant magnetic field through a central bore thereof:

a whole body gradient coil assembly disposed around the central bore;

a radio frequency coil assembly disposed around the central bore;

a head coil removably mounted in the central bore, and head coil including:

a plurality of gradient coil windings disposed on (i) a first cylindrical portion having first and second ends, (ii) a second portion connected to the first cylindrical portion first end and being flared outward to a larger circumference than the first cylindrical portion, and (iii) a third portion connected to the second end of the first cylindrical portion and being flared outward to a larger circumference than the first cylindrical portion, the gradient coil windings being disposed symmetrically on the first, second, and third portions such interactions of the temporally constant magnetic field and electric current pulses in the gradient coil winding induce no net torque on the head coil, and

an RF coil winding disposed within at least the first cylindrical portion;

a magnetic resonance excitation and reconstruction means for controlling the radio frequency and insertable gradient coils for inducing magnetic resonance within the first cylindrical portion and for receiving magnetic resonance signals therefrom and for reconstructing the received magnetic resonance signals into an image representation.

13. The magnetic resonance imaging system as set forth in claim 11 wherein the gradient coil windings of the head coil include axial gradient field windings which extend circumferentially around the first portion and the second portion with a plurality of turns for generating gradient magnetic fields axially along the head gradient coil.

14. The magnetic resonance imaging system as set forth in claim 11 wherein the head gradient coil windings include at least four bunched coils, at least two of the bunched coils each including arc segments extending circumferentially around the first portion and return paths extending along the second circumferential portion.

15. A magnetic resonance imaging system comprising:

a primary magnetic field assembly for generating a temporally constant magnetic field through a central bore thereof;

a whole body gradient coil assembly disposed around the central bore;

a radio frequency coil assembly disposed around the central bore;

head gradient coil removably mounted in the central bore, the head coil including:

at least four thumbprint pattern coil segments which extend at least in part around a cylindrical first portion of a first circumference, at least two of the thumbprint pattern coil segments each extending in part over about one quadrant of the first portion and each extending in part over a second portion of a second circumference, which second circumference is larger than the first circumference, the second circumference portion being mounted toward a patient receiving end of the head gradient coil, and

an RF coil winding disposed within at least the first cylindrical portion;

a magnetic resonance excitation and reconstruction means for controlling the radio frequency and insertable gradient coils for inducing magnetic resonance within the first cylindrical portion and for receiving magnetic resonance signals therefrom and for reconstructing the received magnetic resonance signals into an image representation.

16. The magnetic resonance imaging system as set forth in claim 11 wherein the first portion and the second portion are both circularly symmetric.

17. The magnetic resonance imaging system as set forth in claim 11 wherein the second portion is flared outward from the first portion to define an open, generally conical segment.

18. A gradient coil set for a magnetic resonance imaging system, the gradient coil set comprising:

a gradient coil carrier including (i) a cylinder for defining an imaging region therein, the cylinder having a first end for receiving a patient and a second end and (ii) a flange attached to the first end;

at least two current-carrying continuous paths, each path having a plurality of turns, with each turn being partially disposed on the cylinder and partially on the flange;

a gradient coil power supply for supplying current pulses to the continuous paths for selectively creating a linear magnetic field gradient.

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Main Menu | Search Form | Posting Counts | Show S Numbers | Edit S Numbers | Preferences | Cases

Search Results -

Term	Documents
OPEN.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	1909104
OPENS.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	335235
VERTICAL$3	0
VERTICAL.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	1619248
VERTICALA.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	9
VERTICALALY.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	4
VERTICALAND.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	6
VERTICALARM.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	1
VERTICALBAR.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	1
VERTICALD.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	3
VERTICALE.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	15
(L8 AND (OPEN OR VERTICAL$3)).USPT,PGPB,JPAB,EPAB,DWPI,TDBD.	1

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Database:
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JPO Abstracts Database
EPO Abstracts Database
Derwent World Patents Index
IBM Technical Disclosure Bulletins

Search: L9

Refine Search

Recall Text | Clear

Search History

DATE: Friday, April 04, 2003 Printable Copy Create Case

Set Name side by side	**Query**	**Hit Count**	**Set Name** result set
	DB=USPT,PGPB,JPAB,EPAB,DWPI,TDBD; PLUR=YES; OP=ADJ		
L9	L8 and (open or vertical$3)	1	L9
L8	L7 and (phas$4)	2	L8
L7	L6 and (array$3)	5	L7
L6	L5 and (((radio adj frequency) or RF or radiofrequency or radio-frequency) with coil)	11	L6
L5	L4 and (shoulder)	13	L5
L4	L3 and (shield$4 with coil)	93	L4
L3	L2 and ((uniplanar or uni-planar or flat or planar or smooth) with gradient with coil)	164	L3
L2	L1 and (gradient with coil with (set or plurality or group or multiple))	1313	L2
L1	((magnetic adj resonance) or MRI or NMR)	145092	L1

END OF SEARCH HISTORY

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Search Results -

Term	Documents
CON-JOIN$4	0
CON-JOINED.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	3
CON-JOINING.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	1
CON-JOINT.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	8
CON-JOINTLY.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	10
CONJOIN$4	0
CONJOIN.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	513
CONJOINABLE.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	33
CONJOINABLY.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	1
CONJOINDER.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	18
CONJOINE.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	1
(L2 AND (CON-JOIN$4 OR CONJOIN$4)).USPT,PGPB,JPAB,EPAB,DWPI,TDBD.	0

There are more results than shown above. Click here to view the entire set.

Database:
US Patents Full-Text Database
US Pre-Grant Publication Full-Text Database
JPO Abstracts Database
EPO Abstracts Database
Derwent World Patents Index
IBM Technical Disclosure Bulletins

Search: L16

Refine Search

Recall Text | Clear

Search History

DATE: Friday, April 04, 2003 Printable Copy Create Case

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L15	L3 and (con-join$4 or conjoin$4)	0	L15
L14	L4 and (con-join$4 or conjoin$4)	0	L14
L13	L12 and (phas$4)	2	L13
L12	L10 and (butterfly or helmholz or helmholtz or "figure eight" or figure-eight or saddle or solenoid$4)	6	L12
L11	L10 and (con-join$4 or conjoin$4)	0	L11
L10	L5 and (open or vertical$3)	10	L10
L9	L8 and (open or vertical$3)	1	L9
L8	L7 and (phas$4)	2	L8
L7	L6 and (array$3)	5	L7
L6	L5 and (((radio adj frequency) or RF or radiofrequency or radio-frequency) with coil)	11	L6
L5	L4 and (shoulder)	13	L5
L4	L3 and (shield$4 with coil)	93	L4
L3	L2 and ((uniplanar or uni-planar or flat or planar or smooth) with gradient with coil)	164	L3
L2	L1 and (gradient with coil with (set or plurality or group or multiple))	1313	L2
L1	((magnetic adj resonance) or MRI or NMR)	145092	L1

END OF SEARCH HISTORY

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Search Results - Record(s) 1 through 11 of 11 returned.

1. Document ID: US 20020050895 A1

L6: Entry 1 of 11 File: PGPB May 2, 2002

PGPUB-DOCUMENT-NUMBER: 20020050895
PGPUB-FILING-TYPE: new
DOCUMENT-IDENTIFIER: US 20020050895 A1

TITLE: Magnetic apparatus for MRI

PUBLICATION-DATE: May 2, 2002

INVENTOR-INFORMATION:

NAME	CITY	STATE	COUNTRY	RULE-47
Zuk, Yuval	Haifa		IL	
Katz, Yoav	Rehovot		IL	
Katznelson, Ehud	Ramat Yishai		IL	
Rotem, Haim	Mate Asher		IL	

US-CL-CURRENT: 335/216

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
Draw Desc | Image

2. Document ID: US 6278275 B1

L6: Entry 2 of 11 File: USPT Aug 21, 2001

US-PAT-NO: 6278275
DOCUMENT-IDENTIFIER: US 6278275 B1

TITLE: Gradient coil set with non-zero first gradient field vector derivative

DATE-ISSUED: August 21, 2001

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Petropoulos; Labros S.	Solon	OH		
Schlitt; Heidi A.	Chesterland	OH		

US-CL-CURRENT: 324/318; 324/309, 324/320

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
Draw Desc | Image

3. Document ID: US 6163240 A

L6: Entry 3 of 11 File: USPT Dec 19, 2000

US-PAT-NO: 6163240
DOCUMENT-IDENTIFIER: US 6163240 A

TITLE: Magnetic apparatus for MRI

DATE-ISSUED: December 19, 2000

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Zuk; Yuval	Haifa			IL
Katznelson; Ehud	Ramat Yishai			IL
Katz; Yoav	Rehovot			IL
Rotem; Haim	Mate Asher			IL

US-CL-CURRENT: 335/299; 324/318, 324/319, 324/320, 335/296, 335/302, 335/306

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments KWIC
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4. Document ID: US 5990681 A

L6: Entry 4 of 11 File: USPT Nov 23, 1999

US-PAT-NO: 5990681
DOCUMENT-IDENTIFIER: US 5990681 A

TITLE: Low-cost, snap-in whole-body RF coil with mechanically switchable resonant frequencies

DATE-ISSUED: November 23, 1999

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Richard; Mark A.	S. Euclid	OH		
Mastandrea; Nicholas J.	Euclid	OH		

US-CL-CURRENT: 324/318; 324/319, 324/320, 600/422

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments KWIC
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5. Document ID: US 5952830 A

L6: Entry 5 of 11 File: USPT Sep 14, 1999

US-PAT-NO: 5952830
DOCUMENT-IDENTIFIER: US 5952830 A

TITLE: Octapole magnetic resonance gradient coil system with elongate azimuthal gap

DATE-ISSUED: September 14, 1999

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Petropoulos; Labros S.	Solon	OH		
Mastandrea; Nicholas J.	Euclid	OH		
Richard; Mark A.	South Euclid	OH		

US-CL-CURRENT: 324/318

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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6. Document ID: US 5708360 A

L6: Entry 6 of 11 File: USPT Jan 13, 1998

US-PAT-NO: 5708360
DOCUMENT-IDENTIFIER: US 5708360 A

TITLE: Active shield gradient coil for nuclear magnetic resonance imaging apparatus

DATE-ISSUED: January 13, 1998

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Yui; Masao	Kanagawa-ken			JP
Kondo; Masafumi	Kanagawa-ken			JP

US-CL-CURRENT: 324/318; 324/322

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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7. Document ID: US 5585724 A

L6: Entry 7 of 11 File: USPT Dec 17, 1996

US-PAT-NO: 5585724
DOCUMENT-IDENTIFIER: US 5585724 A

TITLE: Magnetic resonance gradient coils with interstitial gap

DATE-ISSUED: December 17, 1996

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Morich; Michael A.	Mentor	OH		
Petropoulos; Labros S.	Solon	OH		

US-CL-CURRENT: 324/318

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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8. Document ID: US 5581185 A

L6: Entry 8 of 11 File: USPT Dec 3, 1996

US-PAT-NO: 5581185
DOCUMENT-IDENTIFIER: US 5581185 A

TITLE: Torque-balanced gradient coils for magnetic resonance imaging

DATE-ISSUED: December 3, 1996

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Petropoulos; Labros	Cleveland Heights	OH		
Patrick; John L.	Chagrin Falls	OH		
Morich; Michael A.	Mentor	OH		

US-CL-CURRENT: 324/318; 324/322

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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9. Document ID: US 5497089 A

L6: Entry 9 of 11 File: USPT Mar 5, 1996

US-PAT-NO: 5497089
DOCUMENT-IDENTIFIER: US 5497089 A

TITLE: Wide aperture gradient set

DATE-ISSUED: March 5, 1996

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Lampman; David A.	Eastlake	OH		
Morich; Michael A.	Mentor	OH		
Petropoulos; Labros	Cleveland Heights	OH		

US-CL-CURRENT: 324/318; 324/322

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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10. Document ID: US 5485087 A

L6: Entry 10 of 11 File: USPT Jan 16, 1996

US-PAT-NO: 5485087
DOCUMENT-IDENTIFIER: US 5485087 A

TITLE: Magnetic resonance insert gradient coils with parabolic returns for improved access

DATE-ISSUED: January 16, 1996

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Morich; Michael A.	Mentor	OH		
Petropoulos; Labros	Cleveland Heights	OH		
Lampman; David A.	Eastlake	OH		

US-CL-CURRENT: 324/318; 600/422

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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☐ 11. Document ID: US 5278504 A

L6: Entry 11 of 11 File: USPT Jan 11, 1994

US-PAT-NO: 5278504
DOCUMENT-IDENTIFIER: US 5278504 A

TITLE: Gradient coil with off center sweet spot for magnetic resonance imaging

DATE-ISSUED: January 11, 1994

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Patrick; John L.	Chagrin Falls	OH		
Morich; Michael A.	Mentor	OH		
Petropoulos; Labros	Cleveland Hts.	OH		
Hajnal; J. V.	London			GB2
Hall; A. S.	Middlesex			GB2

US-CL-CURRENT: 324/318; 324/322

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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Term	Documents
RADIO.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	392292
RADIOS.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	11955
FREQUENCY.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	1357999
FREQUENCIES.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	285938
FREQUENCYS.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	48
RF.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	174688
RFS.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	930
RADIOFREQUENCY.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	5546
RADIOFREQUENCIES.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	144
RADIOFREQUENCYS	0
RADIO-FREQUENCY.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	14433
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Search Results - Record(s) 1 through 5 of 5 returned.

☐ 1. Document ID: US 6278275 B1

L7: Entry 1 of 5 File: USPT Aug 21, 2001

US-PAT-NO: 6278275
DOCUMENT-IDENTIFIER: US 6278275 B1

TITLE: Gradient coil set with non-zero first gradient field vector derivative

DATE-ISSUED: August 21, 2001

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Petropoulos; Labros S.	Solon	OH		
Schlitt; Heidi A.	Chesterland	OH		

US-CL-CURRENT: 324/318; 324/309, 324/320

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
Draw Desc | Image

☐ 2. Document ID: US 5581185 A

L7: Entry 2 of 5 File: USPT Dec 3, 1996

US-PAT-NO: 5581185
DOCUMENT-IDENTIFIER: US 5581185 A

TITLE: Torque-balanced gradient coils for magnetic resonance imaging

DATE-ISSUED: December 3, 1996

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Petropoulos; Labros	Cleveland Heights	OH		
Patrick; John L.	Chagrin Falls	OH		
Morich; Michael A.	Mentor	OH		

US-CL-CURRENT: 324/318; 324/322

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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☐ 3. Document ID: US 5497089 A

L7: Entry 3 of 5 File: USPT Mar 5, 1996

US-PAT-NO: 5497089
DOCUMENT-IDENTIFIER: US 5497089 A

TITLE: Wide aperture gradient set

DATE-ISSUED: March 5, 1996

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Lampman; David A.	Eastlake	OH		
Morich; Michael A.	Mentor	OH		
Petropoulos; Labros	Cleveland Heights	OH		

US-CL-CURRENT: 324/318; 324/322

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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---

4. Document ID: US 5485087 A

L7: Entry 4 of 5 File: USPT Jan 16, 1996

US-PAT-NO: 5485087
DOCUMENT-IDENTIFIER: US 5485087 A

TITLE: Magnetic resonance insert gradient coils with parabolic returns for improved access

DATE-ISSUED: January 16, 1996

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Morich; Michael A.	Mentor	OH		
Petropoulos; Labros	Cleveland Heights	OH		
Lampman; David A.	Eastlake	OH		

US-CL-CURRENT: 324/318; 600/422

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
Draw Desc | Image

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5. Document ID: US 5278504 A

L7: Entry 5 of 5 File: USPT Jan 11, 1994

US-PAT-NO: 5278504
DOCUMENT-IDENTIFIER: US 5278504 A

TITLE: Gradient coil with off center sweet spot for magnetic resonance imaging

DATE-ISSUED: January 11, 1994

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Patrick; John L.	Chagrin Falls	OH		
Morich; Michael A.	Mentor	OH		
Petropoulos; Labros	Cleveland Hts.	OH		
Hajnal; J. V.	London			GB2
Hall; A. S.	Middlesex			GB2

US-CL-CURRENT: 324/318; 324/322

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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Term	Documents
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ARRAY.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	521991
ARRAYA.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	8
ARRAYAL.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	271
ARRAYALS.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	20
ARRAYAM.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	1
ARRAYAN.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	2
ARRAYAND.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	6
ARRAYARE.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	1
ARRAYAS.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	4
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Search Results - Record(s) 1 through 2 of 2 returned.

☐ 1. Document ID: US 6278275 B1

L8: Entry 1 of 2 File: USPT Aug 21, 2001

US-PAT-NO: 6278275
DOCUMENT-IDENTIFIER: US 6278275 B1

TITLE: Gradient coil set with non-zero first gradient field vector derivative

DATE-ISSUED: August 21, 2001

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Petropoulos; Labros S.	Solon	OH		
Schlitt; Heidi A.	Chesterland	OH		

US-CL-CURRENT: 324/318; 324/309, 324/320

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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☐ 2. Document ID: US 5278504 A

L8: Entry 2 of 2 File: USPT Jan 11, 1994

US-PAT-NO: 5278504
DOCUMENT-IDENTIFIER: US 5278504 A

TITLE: Gradient coil with off center sweet spot for magnetic resonance imaging

DATE-ISSUED: January 11, 1994

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Patrick; John L.	Chagrin Falls	OH		
Morich; Michael A.	Mentor	OH		
Petropoulos; Labros	Cleveland Hts.	OH		
Hajnal; J. V.	London			GB2
Hall; A. S.	Middlesex			GB2

US-CL-CURRENT: 324/318; 324/322

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
Draw Desc | Image

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Term	Documents
PHAS$4	0
PHAS.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	770
PHASA.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	10
PHASABLE.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	6
PHASACC.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	1
PHASACID.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	1
PHASAE.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	19
PHASAGE.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	3
PHASAL.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	275
PHASALLY.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	27
PHASAND.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	1
(L7 AND (PHAS$4)).USPT,PGPB,JPAB,EPAB,DWPI,TDBD.	2

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Search Results - Record(s) 1 through 1 of 1 returned.

☐ 1. Document ID: US 6278275 B1

L9: Entry 1 of 1 File: USPT Aug 21, 2001

US-PAT-NO: 6278275
DOCUMENT-IDENTIFIER: US 6278275 B1

TITLE: Gradient coil set with non-zero first gradient field vector derivative

DATE-ISSUED: August 21, 2001

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Petropoulos; Labros S.	Solon	OH		
Schlitt; Heidi A.	Chesterland	OH		

US-CL-CURRENT: 324/318; 324/309, 324/320

Full | Title | CIT.1 | REV.1 | CLS.1 | REF.1 | SEQ.1 | ATT.1 | KWIC | RAW.1

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Term	Documents
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OPENS.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	335235
VERTICAL$3	0
VERTICAL.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	1619248
VERTICALA.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	9
VERTICALALY.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	4
VERTICALAND.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	6
VERTICALARM.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	1
VERTICALBAR.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	1
VERTICALD.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	3
VERTICALE.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	15
(L8 AND (OPEN OR VERTICAL$3)).USPT,PGPB,JPAB,EPAB,DWPI,TDBD.	1

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L9: Entry 1 of 1 File: USPT Aug 21, 2001

DOCUMENT-IDENTIFIER: US 6278275 B1
TITLE: Gradient coil set with non-zero first gradient field vector derivative

Abstract Text (1):
A gradient coil assembly (22) generates substantially linear magnetic gradients across the central portion of an examination region (14). The gradient coil assembly (22) includes primary x, y, and z-gradient coils (62, 66, 68) which generate a gradient magnetic field (90) having a non-zero first derivative in and adjacent the examination region. Preferably, the gradient coil assembly (22) includes secondary, shielding x, y, and z coils which generate a magnetic field which substantially cancels, in an area outside a region defined by the shielding coils, a fringe magnetic field generated by the primary gradient coils. The existence of a non-zero first derivative in and adjacent the examination region eliminates aliasing effects attributable to the non-unique gradient field values on either side of a rollover point (82). The non-unique values of the gradient magnetic field adjacent the rollover point caused structure near the rollover point to overlay each other (FIGS. 7B, 8B). The unique non-linearity of the present gradient (90) adjacent the edges expands (magnifies) the image adjacent the edges (FIGS. 7A, 8A). Because the expansion is unique, distortions at the edges are readily and accurately mapped (52) back to linear.

Brief Summary Text (2):
The present invention relates to the magnetic resonance arts. It finds particular application in conjunction with gradient coils for a magnetic resonance imaging apparatus and will be described with particular reference thereto. However, it is to be appreciated that the present invention will also find application in conjunction with localized magnetic resonance spectroscopy systems and other applications which utilize gradient magnetic fields.

Brief Summary Text (3):
In magnetic resonance imaging, a uniform magnetic field is created through an examination region in which a subject to be examined is disposed. A series of radio frequency pulses and magnetic field gradients are applied to the examination region. Gradient fields are conventionally applied as a series of gradient pulses with pre-selected profiles. The radio frequency pulses excite magnetic resonance and the gradient field pulses phase and frequency encode the induced resonance. In this manner, phase and frequency encoded magnetic resonance signals are generated.

Brief Summary Text (4):
More specifically, the gradient magnetic field pulses are typically applied to select and encode the magnetic resonance with spatial position. In some embodiments, the magnetic field gradients are applied to select a slice or slab to be imaged. Ideally, the phase or frequency encoding uniquely identifies spatial location.

Brief Summary Text (6):
Historically, gradient coil designs were developed in a "forward approach," whereby a set of initial coil positions were defined and the fields, energy, and inductance calculated. If these quantities were not within the particular design criteria, the coil positions were shifted (statistically or otherwise) and the results re-evaluated. This iterative procedure continued until a suitable design was obtained.

Brief Summary Text (7):
Recently, gradient coils are designed using the "inverse approach," whereby gradient

fields are forced to match predetermined values at specified spatial locations inside the imaging volume. Then, a continuous current density is generated which is capable of producing such fields. This approach is adequate for designing non-shielded or actively shielded gradient coil sets.

Brief Summary Text (8):
Often, shielded gradient coil sets are designed such that their gradient magnetic field has an inherent rollover point along, but near the outer edge of its perspective gradient axis. That is, the first derivative of the gradient magnetic field is zero at a certain location along the gradient axis and inside the physical volume bounded by the inner surface of the gradient structure. The gradient magnetic field takes on non-unique values after passing the rollover point where the first derivative of the gradient magnetic field is zero. The rollover point may be in the center or near the edge of the bore, beyond where the subject is positioned. This design is problematic for an imaging sequence with a large field of view (FOV) in which portions of the subject are disposed between the rollover point and the bore. Areas of a subject that are located beyond the rollover point will alias back into the image, which causes ghosting and distortion of the image. A gradient deghosting and distortion algorithm is then implemented during postprocessing in order to compensate for distortions in the image. The gradient distortion algorithm, particularly when applied to all three gradient coils, extends the image postprocessing time and extends significantly the overall time of the magnetic resonance study.

Brief Summary Text (9):
In addition, information in the raw data related to the voxels located beyond the rollover point cannot be recovered uniquely. Voxels on either side of the rollover point that experience the same gradient strength are encoded indistinguishably. This limits the maximum FOV of a given sequence and limits the range of translational movement for the examined subject inside the image volume. This problem is particularly apparent when imaging extremities, such as shoulder, wrists, legs, and elbows, because typically these regions are located near the rollover point. Therefore, any attempt to move one side of an extremity near the isocenter of the imaging volume places the other side in the vicinity of the rollover point, which results in the aforementioned aliasing problems.

Brief Summary Text (10):
The present invention contemplates a new and improved gradient coil set which overcomes the above-referenced problems and others.

Brief Summary Text (12):
In accordance with one aspect of the present invention, a magnetic resonance imaging apparatus includes a main magnet for generating a main magnetic field through and surrounding an examination region. A gradient coil assembly generates gradient magnetic fields across the examination region. The gradient magnetic fields have a non-zero first derivative in and adjacent the examination region. An RF transmitter and coil assembly positioned adjacent the examination region excites magnetic resonance dipoles in and adjacent the examination region. An RF coil and receiver assembly receives and demodulates magnetic resonance signals from the resonating dipoles. A reconstruction processor reconstructs the demodulated magnetic resonance signals into an image representation.

Brief Summary Text (13):
In accordance with another aspect of the present invention, a method of magnetic resonance imaging includes inducing resonance in selected dipoles in an examination region such that the selected dipoles generate magnetic resonance signals. A gradient magnetic field is applied across the examination region to encode the magnetic resonance signals along at least one axis. The gradient magnetic field has a non-zero first derivative through and adjacent edges of the examination region. Further, the encoded magnetic resonance signals are received and demodulated. Finally, the demodulated resonance signals are reconstructed into an image representation.

Brief Summary Text (14):
In accordance with another aspect of the present invention, a method of designing a gradient coil assembly for a magnetic resonance imaging system includes selecting radius and length values for a primary gradient coil set and a secondary shielding coil set. The method further includes generating a first continuous current distribution for the primary gradient coil set. The first continuous current

distribution is confined within predetermined finite geometric boundaries of a first surface defined above. The first continuous current distribution generates a gradient magnetic field across an examination region where the first derivative of the gradient magnetic field in and adjacent the examination region is non-zero. Further, a second continuous current distribution is generated for the secondary, shielding coil set. The second continuous current distribution is confined within the predetermined finite geometric boundaries defined above. The first and second continuous current distributions generate a magnetic field which substantially cancels in an area outside the region defined by the secondary, shielding coil set. Next, the primary gradient coil set with the secondary, shielding coil set are optimized using an energy/inductance minimization algorithm. Finally, the primary gradient coil set and secondary, shielding coil set are discretized.

Brief Summary Text (15):
In accordance with another aspect of the present invention, a gradient coil assembly for generating magnetic gradients across a main magnetic field of a magnetic resonance apparatus includes x and y-gradient coils which are configured to generate magnetic field gradients across an examination region along first and second orthogonal axes. The first derivative of the magnetic gradient field generated by the x and y-gradient coils is non-zero in and adjacent the examination region. A z-gradient coil generates magnetic field gradients along a third axis which is orthogonal to the first and second axes. The first derivative of the magnetic field gradient generated by the z-gradient coil is non-zero in and adjacent the examination region.

Brief Summary Text (16):
One advantage of the present invention is that it eliminates aliasing effects for magnetic resonance sequences with large fields of view.

Brief Summary Text (18):
Another advantage of the present invention is that it reduces postprocessing time for a magnetic resonance image.

Brief Summary Text (19):
Another advantage of the present invention is that it reduces overall time for a magnetic resonance study.

Drawing Description Text (3):
FIG. 1 is a diagrammatic illustration of a magnetic resonance imaging system in accordance with the present invention;

Drawing Description Text (5):
FIG. 2B is a perspective view of a primary gradient coil set in accordance with the present invention;

Drawing Description Text (7):
FIG. 4 is a flow chart for designing a shielded gradient coil assembly with a non-zero first derivative of the gradient magnetic field in accordance with the present invention;

Drawing Description Text (8):
FIGS. 5A and 5B are diagrammatic illustrations of one quadrant of an exemplary primary x-gradient coil and secondary shielding coil in accordance with the present invention;

Drawing Description Text (9):
FIGS. 6A and 6B are diagrammatic illustrations of one quadrant of an exemplary primary y-gradient coil and secondary shielding coil in accordance with the present invention;

Drawing Description Text (10):
FIG. 7A is a distortion grid for a transverse slice through the z=0.0 plane for an exemplary gradient coil set with no rollover point in accordance with the present invention;

Drawing Description Text (11):
FIG. 7B is a distortion grid for a transverse slice through the z=0.0 plane for an exemplary gradient coil set with an inherent rollover point in accordance with the prior art;

Drawing Description Text (12):
FIG. 8A is a coronal distortion grid for an exemplary gradient coil set with no rollover point in accordance with the present invention; and

Drawing Description Text (13):
FIG. 8B is a coronal distortion grid an exemplary gradient coil set with an inherent rollover point in accordance with the prior art.

Detailed Description Text (2):
With reference to FIG. 1, a main magnetic field control 10 controls superconducting or resistive magnets 12 such that a substantially uniform, temporally constant main magnetic field is created along a z axis through an examination region 14. Although a bore-type magnet is illustrated in FIG. 1, it is to be appreciated that the present invention is equally applicable to open magnetic systems with vertically directed fields. A couch (not illustrated) suspends a subject to be examined within the examination region 14. A magnetic resonance echo means applies a series of radio frequency (RF) and magnetic field gradient pulses to invert or excite magnetic spins, induce magnetic resonance, refocus magnetic resonance, manipulate magnetic resonance, spatially and otherwise encode the magnetic resonance, to saturate spins, and the like to generate magnetic resonance imaging and spectroscopy sequences. More specifically, gradient pulse amplifiers 20 apply current pulses to a gradient coil assembly 22 that includes pairs of primary gradient coil assemblies 22a and shield gradient coil assemblies 22b with no roll-over point to create magnetic field gradients along x, y, and z axes of the examination region 14 with zero or minimal fringe fields outside of the bore. A digital radio frequency transmitter 24 transmits radio frequency pulses or pulse packets to a whole-body RF coil 26 to transmit RF pulses into the examination region 14. A typical radio frequency pulse is composed of a packet of immediately contiguous pulse segments of short duration which, taken together with each other and any applied gradients, achieve a selected magnetic resonance manipulation. For whole-body applications, the resonance signals are commonly picked up by the whole-body RF coil 26, but may be picked up by other specialized RF coils.

Detailed Description Text (3):
For generating images of local regions of the subject, specialized radio frequency coils are placed contiguous to the selected region. For example, an insertable RF coil may be inserted surrounding a selected region at the isocenter of the bore. The insertable RF coil is used to excite magnetic resonance and receive magnetic resonance signals emitting from the patient in the region being examined. Alternatively, the insertable RF coil can be used only to receive resonance signals introduced by whole-body RF coil transmissions. The resultant radio frequency signals are picked up by the whole-body RF coil 26, the insertable RF coil, or other specialized RF coils and demodulated by a receiver 30, preferably a digital receiver.

Detailed Description Text (4):
A sequence control circuit 40 controls the gradient pulse amplifiers 20 and the transmitter 24 to generate any of a plurality of multiple echo sequences such as echo planar imaging, echo volume imaging, gradient and spin echo imaging, fast spin echo imaging, and the like. For the selected sequence, the receiver 30 receives a plurality of data lines in rapid succession following each RF excitation pulse. Ultimately, the radio frequency signals received are demodulated and reconstructed into an image representation by a reconstruction processor 50 which applies a two-dimensional Fourier transform or other appropriate reconstruction algorithm. The image is then stored in an image memory 54. As explained below, the resultant image adjacent its edges tends to be distorted (stretched or contracted). An image linearity correction processor 52 corrects the non-linearity. For example, the gradient field distortion can be empirically measured (see below) and the image can be mapped with the inverse of the mapped field distortion. Other distortion correction algorithms, as are known in the art, can also be utilized. Optionally, the distortion correction can be made in Fourier space prior to reconstruction, incorporated into the reconstruction algorithm, or downstream from the image memory. A human-readable display 56, such as a video monitor, provides a human-readable display of the resultant image. The image may represent a planar slice through the patient, an array of parallel planar slices, a three-dimensional volume, or the like.

Detailed Description Text (6):

A primary z-gradient coil is also constructed of a conductive material, such as foil or wire. The primary z-gradient coil is preferably wound into groves in the inner former 60 and potted in an epoxy. The secondary gradient coil assembly 22b also includes an outer dielectric former 70 of radius b. The secondary x, y, and z shielding coils (not shown) are laminated into the cylindrical surface of the outer former 70 or on longitudinal rods, known as "combs," analogous to the primary gradient coils. These shielding coils are designed to cooperate with the primary gradient coils to generate a magnetic field which has a substantially zero magnetic flux density outside an area defined by the outer former.

Detailed Description Text (7):
With reference to FIG. 3, prior art gradient coils sets typically are designed such that their gradient magnetic field profile 80 has an inherent rollover point 82 along, but near the edge of its respective axis, as shown. At the rollover point, the first derivative of the gradient magnetic field is zero. After passing the rollover point, where the first derivative is zero, the gradient field takes on non-unique values, i.e., assumes identical values to the gradient field on both sides of the rollover point. This leads to aliasing. When portions of the subject are disposed between the rollover point and the bore, areas of the subject that are located beyond the rollover point will alias back into the image, which causes ghosting of the image. Signals from two planes near the edge that are subject to the same gradient field strength are indistinguishable and are combined. In this manner, a ghost of the material beyond the rollover point is folded back on the material inside the rollover point.

Detailed Description Text (9):
The theoretical development, the design procedure and the numerical results for an exemplary shielded gradient coil with no rollover point of the gradient magnetic field along its perspective axis and inside the physical boundaries defined by the inner surface of the gradient tube is now discussed. Specifically, the theoretical development, the design, and the results of a gradient coil where the z component of the magnetic field varies linearly along the transverse direction (x, y-gradient coil), as well as, the axial gradient coil (z-gradient coil) will be presented. The x-gradient coil will be presented in its entirety as a representative for the transverse coils.

Detailed Description Text (10):
The flow chart for designing such a gradient coil structure is shown in FIG. 4. Initially, a geometric configurations of the primary gradient coil step 100 sets the primary coil configuration and a secondary shielding coil configuration step 102 sets the secondary coil configuration. Namely, radius and length for each coil set are chosen. Next, an energy/inductance minimization step 104 optimizes the primary gradient coil set. As a result of the minimization process 104, a first continuous current distribution generation step 106 generates the current distribution for the primary gradient coil set. The first continuous current distribution is confined to the geometric boundaries defined in step 100. The first current distribution is selected such that it generates a magnetic gradient field across the examination region where the first derivative of the gradient magnetic field in and around the examination region is non-zero. Following this step, a second continuous current distribution selection step 108 generates the current distribution for the secondary, shielding coil set such that the second continuous current distribution is confined to the geometric boundaries defined in step 102. The second continuous current distribution generates a magnetic field which, when combined with the magnetic field from the first current distribution, generates a substantially zero fringe magnetic field outside the secondary coil.

Detailed Description Text (11):
Further, in a current discretization step 110, the continuous current distribution of the primary gradient coil set and the secondary, shielding coil set are discretized to generate the number of turns which is required for each coil within each coil set. Optionally, a verifying step 112 applies the Biot-Savart law to the discrete current pattern to verify its validity. Finally, in a measuring and mapping step 114, non-linearities present in the gradient magnetic field near the edges of the examination region are measured and mapped back in order to correct the image near the edges.

Detailed Description Text (13):
The design of a finite, shielded transverse x-gradient coil involves the design of the primary coil (the coil closest to the subject) based on the inverse approach

methodology. For the exemplary transverse coil the gradient magnetic field is anti-symmetric in the x direction around the geometric center of the coil, while it is symmetric along the y and z directions. To generate such a field, the analytical expression of the current for the primary coil Ja(r) can be written as:

Detailed Description Text (16):
In order minimize the fringe field of the primary coil in the area which is outside both the primary and the shielding coils, the Fourier transform of the current for the shielding coil satisfies the following relationship: ##EQU2##

Detailed Description Text (28):
Inverting the previous matrix equation, a solution for j.sup.a.sub..phi.n, and hence for the current density, is obtained. When the continuous current distribution for both the primary and shield coils is evaluated, the stream function technique is used to discretize the current density for both primary and shield coils in such a way that the absolute integer number of turns is obtained for both coils for a given common current value per loop. The discretization and the magnetic gradient field inside the desired imaging volume are then calculated proceeding with steps 6 through 8 of FIG. 4.

Detailed Description Text (29):
For the design of the exemplary primary x-gradient coil, the radius of the cylinder for the primary coil is set equal to 0.3438500 m and its total length is restricted to 1.066400 m. In addition, the radius of the secondary coil is equal to 0.435224 m. The constraints for the design of the primary coil are shown in Table 1. The constraints for the primary y-gradient coil are shown in Table 2.

Detailed Description Text (30):
As shown in Table 1, the first constraint point defines a gradient strength for the first primary and single shield coil to be 27.0 mT/m, the second constraint point specifies a +0.1% linearity of the gradient field along the gradient (x) axis and up to the distance of 23.0 cm for the isocenter of the gradient field, while the third constraint point specifies a -20% uniformity of the gradient field inside the 40 cm imaging volume.

Detailed Description Text (31):
For the exemplary y-gradient shielded coil, the radius of the primary coil is a =0.336040 m with a length of 1.0534 m, while the radius of the secondary coil is b=0.431414 m. As shown in Table 2, the first constraint point defines a gradient strength to be 27.0 mT/m, the second constraint point specifies a +10% linearity of the gradient field along the gradient (x) axis and up to the distance of 26.5 cm for the isocenter of the gradient field, while the third constraint point specifies a -20% uniformity of the gradient field inside the 40 cm imaging volume.

Detailed Description Text (32):
With the presence of these constraints on Tables 1 and 2, and the application of the inverse approach methodology of FIG. 4, the values for the Fourier coefficients for the current density of the shielded x and y-gradient coils are generated. Applying the Stream Function technique to the continuous current densities for both transverse shielded coils, the discrete current patterns for these coils were generated. Specifically, the x-gradient coil, the Stream Function technique generates 23 discrete loops on one quadrant of the primary coil, as shown in FIGS. 5A and 11 loops on one quadrant of the single shield, as shown in FIG. 5B. The common current per loop is 376.99 amps. In this case, the eddy current from the discrete coil configuration is 0.245% over a 50 cm DSV.

Detailed Description Text (33):
By discretizing the current density for the y-gradient coil, the current density for one quadrant of the exemplary primary coil is approximated by 23 loops with a common current of 375.11 amps, as shown in FIG. 6A, while the one quadrant of the shielding coil can be approximated by 10 loops carrying the same current per loop (FIG. 6B). For the y-gradient coil, the eddy currents are only 0.257%. Employing the Biot-Savart law to the discrete current densities for both the x- and y shielded gradient coils, the gradient magnetic field for both of these coils is evaluated along the perspective gradient axis and at the z=0.0 plane. The behavior of the gradient magnetic field for the x-gradient coil again is substantially as illustrated in FIG. 3.

Detailed Description Text (34):

The behavior of the gradient magnetic field for the y-gradient coil is substantially as shown in FIG. 3. Table 3 illustrates the magnetic properties for the x and y shielded gradient coils in more specific detail with a single shield.

Detailed Description Text (35):
Initially, the design of the finite shielded axial z-gradient coil involves the design of the primary coil (the coil that is closest to the subject) based on the inverse approach methodology. For z-gradient coil, the gradient magnetic field is anti-symmetric in the z direction around the geometric center of the coil, while it is symmetric along the x and y directions. Thus, in this case there is no azimuthal dependance on the current density. To generate such a field, the analytical expression of the current for the primary coil J.sup.a (r) can be written as:

Detailed Description Text (38):
In order to minimize the fringe field of the primary coil in the area which is outside both the primary and the shielding coil, the Fourier transform of the current for the shielding coil satisfies the following relationship: ##EQU10##

Detailed Description Text (50):
Inverting the previous matrix equation, a solution for j.sup.a.sub..phi.n, and hence for the current density, is obtained. When the continuous current distribution for both the primary and shield coils J.sup.a, J.sup.b is evaluated, the application of the center of mass technique yields the discrete loop patterns for both primary and shield coils with the extra constraint that the absolute integer number of turns for both coils for a given common current value per loop is obtained. The discretization and the magnetic gradient field inside the desired imaging volume are then calculated proceeding with steps 6 through 8 of FIG. 4.

Detailed Description Text (52):
As shown in Table 4, the first constraint point defines a gradient strength for the primary and shield coil to be 25.0 mT/m, the second constraint point specifies a +10% linearity of the gradient field along the gradient (z) axis and up to the distance of 26.5 cm for the isocenter of the gradient field, while the rest of the constraint points specify the uniformity of the gradient field inside the 45 cm imaging volume.

Detailed Description Text (53):
With the presence of these constraints on Table 4 and the application of the inverse approach methodology of FIG. 4, the values for the Fourier coefficients for the current density of the shielded z-gradient coil are generated. Applying the center of mass technique to the continuous current densities for both the primary coil and the shielding coil, the discrete current patterns for these coils were generated. Specifically, for the preferred first primary and the shield configuration, the center of mass technique generates 60 discrete loops on the primary coil and 30 loops on the single shield. The common current per loop is 347.388 amps. In this case, the eddy current from the discrete coil configuration is 0.19% over a 50 cm DSV. Table 5 illustrates the magnetic properties of the shielded x-gradient coil.

Detailed Description Text (58):
The present invention is applicable to other alternative gradient coil geometries, such as elliptical, planar, flared, etc., as well as the asymmetric gradient coil designs or any combination thereof. The present invention is also applicable to the design of gradient coil structures suitable for vertically oriented or open magnet systems. Further, the disclosed primary and screen coil set can be bunched (concentrated) or thumbprint designs generated using forward or inverse approach methods. In addition, the primary and the shield coils can have any possible mixing of bunched and/or thumbprint designs. It is contemplated that zero net thrust force or torque can be incorporated into the proposed design algorithm in a known manner.

Detailed Description Paragraph Table (3):
TABLE 3 Gradient field characteristics for the shield x and y-gradient coils. Properties x-gradient coil y-gradient coil Gradient Strength 27 27 (mT/m) Gradient Linearity 0.8% 7.8% (.rho. = .+-.22.5 Cm) Gradient Uniformity -20% -20% (z = .+-.20.0 Cm) Rise Time @ 700V 465 .mu.sec Slew Rate @ 700V 60 T/m/sec 61 T/m/sec % Eddy Current on 0.245% 0.257% 50 cm DSV

CLAIMS:

1. A magnetic resonance imaging apparatus comprising:

a main magnet for generating a main magnetic field through and surrounding an examination region;

a gradient coil assembly for generating gradient magnetic fields across the examination region, which gradient magnetic fields have a non-zero first derivative in and adjacent the examination region, the gradient coil assembly including:

a primary gradient coil set disposed adjacent the examination region, said primary gradient coil set including an array of conductive loops for generating the gradient magnetic fields along three orthogonal axes;

a secondary shielding coil set disposed between the primary coil assembly and the main magnet, said secondary shielding coil set including an array of conductive loops such that a current density flowing thereon causes a magnetic flux density which interacts with a magnetic flux density generated by the primary magnetic field to substantially zero a net magnetization flux density outside an area defined by the secondary shielding coil set;

an RF transmitter and coil assembly positioned adjacent the examination region such that it excites magnetic resonance dipoles in and adjacent the examination region;

an RF coil and receiver assembly which receives and demodulates magnetic resonance signals from the resonating dipoles; and

a reconstruction processor for reconstructing the demodulated magnetic resonance signals into an image representation.

2. The magnetic resonance imaging apparatus according to claim 1, wherein primary gradient coil set and secondary shielding coil set are arranged on formers.

3. The magnetic resonance imaging apparatus according to claim 2, wherein the formers are hollow cylindrical tubes arranged such that the examination region is defined inside the former of the primary gradient coil set with the former of the primary gradient coil set positioned inside the former of the secondary, shielding coil set.

4. A magnetic resonance imaging apparatus comprising:

a main magnet for generating a main magnetic field through and surrounding an examination region;

a gradient coil assembly for generating gradient magnetic fields across the examination region, a gradient magnetic field generated along at least one axis having (i) a substantially constant slope along a central region of the examination region and (ii) an increasingly step slope adjacent edges of the examination region;

an RF transmitter and coil assembly positioned adjacent the examination region such that it excites magnetic resonance dipoles in and adjacent the examination region;

an RF coil and receiver assembly which receives and demodulates magnetic resonance signals from the resonating dipoles;

a reconstruction processor for reconstructing the demodulated magnetic resonance signals into an image representation; and

a linearity correction processor which adjusts at least one of the demodulated resonance signals and the image representation to correct for distortion attributable to the increasingly steep slope of the gradient magnetic field adjacent edges of the examination region.

5. A magnetic resonance imaging apparatus comprising:

a main magnet for generating a main magnetic field through and surrounding an examination region;

a gradient coil assembly for generating gradient magnetic fields across the examination region, the gradient coil assembly including:

three primary gradient coil sets, one for generating a gradient magnetic field along each of three orthogonal axes, each of the primary gradient coil sets generating a corresponding gradient magnetic field which is linear adjacent a central region of the examination region and monotonically increasing adjacent edges of the examination region;

an RF transmitter and coil assembly positioned adjacent the examination region such that it excites magnetic resonance dipoles in and adjacent the examination region;

an RF coil and receiver assembly which receives and demodulates magnetic resonance signals from the resonating dipoles; and

a reconstruction processor for reconstructing the demodulated magnetic resonance signals into an image representation.

6. The magnetic resonance imaging apparatus according to claim 5, further including a secondary coil disposed around the primary gradient coil sets, the primary gradient coil sets and the secondary coil cooperating to generate the magnetic field gradients in the examination region and to minimize magnetic field gradients outside of the examination region.

7. A method of magnetic resonance imaging comprising:

inducing resonance in selected dipoles in an examination region such that the selected dipoles generate magnetic resonance signals;

applying a gradient magnetic field along three orthogonal axes across the examination region to encode the magnetic resonance signals, a first derivative of the gradient magnetic field being non-zero throughout the examination region, such that the gradient magnetic field along each axis is unique in and adjacent edges of the examination region;

receiving and demodulating the encoded resonance signals;

reconstructing the demodulated resonance signals into an image representation.

8. A magnetic resonance imaging method comprising:

inducing resonance in selected dipoles in an examination region such that the selected dipoles generate magnetic resonance signals;

applying a gradient magnetic field across the examination region to encode the magnetic resonance signals along at least one axis, the gradient magnetic field (i) having a non-zero first derivative through the examination region, (ii) being substantially linear across a central region of the examination region, and (iii) changing strength monotonically adjacent edges of the examination region;

receiving and demodulating the encoded resonance signals;

reconstructing the demodulated resonance signals into an image representation.

9. A magnetic resonance imaging method comprising:

inducing resonance in selected dipoles in an examination region such that the selected dipoles generate magnetic resonance signals;

applying a gradient magnetic field across the examination region to encode the magnetic resonance signals along at least one axis, the gradient magnetic field having (i) a substantially constant slope across the central region of the examination region and (ii) a continuously increasing slope adjacent an edge of the examination region;

receiving and demodulating the encoded resonance signals;

reconstructing the demodulated resonance signals into an image representation; and

adjusting one of (i) the demodulated resonance signals and (ii) the reconstructed image representation to correct for distortions attributable to the continuously

increasing slope of the gradient magnetic field adjacent the edge of the examination region.

10. A method of designing a gradient coil assembly for magnetic resonance imaging systems, the method comprising:

(a) selecting radius and length for a primary gradient coil set and radius and length for a secondary, shielding coil set;

(b) generating a first continuous current distribution for the primary gradient coil set such that the first continuous current distribution is confined within predetermined finite geometric boundaries of a first surface defined in step (a), said first continuous current distribution generating a gradient magnetic field across an examination region whose first derivative in and adjacent the examination region is non-zero;

(c) generating a second continuous current distribution for the secondary, shielding coil set such that the second continuous current distribution is confined within the predetermined finite geometric boundaries defined in step (a), the first and second continuous current distributions generating a magnetic field which substantially cancels in an area outside a region defined by the secondary, shielding coil set;

(d) optimizing the primary gradient coil set with the secondary, shielding coil set utilizing an energy/inductance minimization algorithm; and

(e) discretizing the primary gradient coil set and the secondary, shielding coil set.

11. The method according to claim 10, wherein the method further comprises:

(f) applying the Biot-Savart law to the discrete current pattern to verify its validity; and

(g) measuring and mapping non-linearities present in one of (i) the gradient magnetic field near edges of the examination region and (ii) edges of magnetic resonance images of a subject extending substantially to the edges of the examination region in order to generate a correction map.

12. A shielded gradient coil assembly designed by the method of claim 12.

13. A gradient coil assembly for generating magnetic gradients across a main magnetic field of a magnetic resonance apparatus, the gradient coil assembly comprising:

x and y-gradient coils configured to generate magnetic field gradients across an examination region along first and second orthogonal axes, a first derivative of the magnetic gradient field generated by the x and y-gradient coils in and adjacent the examination region being non-zero; and

a z-gradient coil for generating magnetic field gradients along a third axis orthogonal to the first and second axes, a first derivative of the magnetic field gradient generated by the z-gradient coil in and adjacent the examination region being non-zero.

16. A magnetic resonance imaging apparatus comprising:

a main magnet for generating a main magnetic field through and surrounding an examination region;

a gradient coil assembly for generating gradient magnetic fields along three orthogonal axes across the examination region, a first derivative of the gradient magnetic fields generated by the gradient coil assembly being non-zero in and adjacent the examination region;

an RF transmitter and coil assembly positioned adjacent the examination region such that it excites magnetic resonance dipoles in and adjacent the examination region;

an RF coil and receiver assembly which receives and demodulates magnetic resonance signals from the resonating dipoles; and

a reconstruction processor for reconstructing the demodulated magnetic resonance signals into an image representation.

17. The magnetic resonance imaging apparatus according to claim 16, wherein the gradient magnetic field generating along at least one axis has (i) a substantially constant slope along a central region of the examination region and (ii) an increasingly steep slope adjacent edges of the examination region.

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☐ 1. Document ID: US 20020050895 A1

L10: Entry 1 of 10 File: PGPB May 2, 2002

PGPUB-DOCUMENT-NUMBER: 20020050895
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DOCUMENT-IDENTIFIER: US 20020050895 A1

TITLE: Magnetic apparatus for MRI

PUBLICATION-DATE: May 2, 2002

INVENTOR-INFORMATION:

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US-CL-CURRENT: 335/216

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | Claims | KWIC | Draw. Desc | Image

☐ 2. Document ID: US 6297635 B1

L10: Entry 2 of 10 File: USPT Oct 2, 2001

US-PAT-NO: 6297635
DOCUMENT-IDENTIFIER: US 6297635 B1

TITLE: Switchable gradient coil arrangement

DATE-ISSUED: October 2, 2001

INVENTOR-INFORMATION:

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US-CL-CURRENT: 324/318; 324/322

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | Claims | KWIC | Draw. Desc | Image

☑ 3. Document ID: US 6278275 B1

L10: Entry 3 of 10 File: USPT Aug 21, 2001

US-PAT-NO: 6278275
DOCUMENT-IDENTIFIER: US 6278275 B1

TITLE: Gradient coil set with non-zero first gradient field vector derivative

DATE-ISSUED: August 21, 2001

INVENTOR-INFORMATION:

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US-CL-CURRENT: 324/318; 324/309, 324/320

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | Claims | KWIC
Draw. Desc | Image

☐ 4. Document ID: US 6163240 A

L10: Entry 4 of 10 File: USPT Dec 19, 2000

US-PAT-NO: 6163240
DOCUMENT-IDENTIFIER: US 6163240 A

TITLE: Magnetic apparatus for MRI

DATE-ISSUED: December 19, 2000

INVENTOR-INFORMATION:

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US-CL-CURRENT: 335/299; 324/318, 324/319, 324/320, 335/296, 335/302, 335/306

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
Draw. Desc | Image

☑ 5. Document ID: US 5990681 A

L10: Entry 5 of 10 File: USPT Nov 23, 1999

US-PAT-NO: 5990681
DOCUMENT-IDENTIFIER: US 5990681 A

TITLE: Low-cost, snap-in whole-body RF coil with mechanically switchable resonant frequencies

DATE-ISSUED: November 23, 1999

INVENTOR-INFORMATION:

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US-CL-CURRENT: 324/318; 324/319, 324/320, 600/422

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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6. Document ID: US 5952830 A

L10: Entry 6 of 10 File: USPT Sep 14, 1999

US-PAT-NO: 5952830
DOCUMENT-IDENTIFIER: US 5952830 A

TITLE: Octapole magnetic resonance gradient coil system with elongate azimuthal gap

DATE-ISSUED: September 14, 1999

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
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Richard; Mark A.	South Euclid	OH		

US-CL-CURRENT: 324/318

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
Draw Desc | Image

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7. Document ID: US 5708360 A

L10: Entry 7 of 10 File: USPT Jan 13, 1998

US-PAT-NO: 5708360
DOCUMENT-IDENTIFIER: US 5708360 A

TITLE: Active shield gradient coil for nuclear magnetic resonance imaging apparatus

DATE-ISSUED: January 13, 1998

INVENTOR-INFORMATION:

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US-CL-CURRENT: 324/318; 324/322

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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8. Document ID: US 5585724 A

L10: Entry 8 of 10 File: USPT Dec 17, 1996

US-PAT-NO: 5585724
DOCUMENT-IDENTIFIER: US 5585724 A

TITLE: Magnetic resonance gradient coils with interstitial gap

DATE-ISSUED: December 17, 1996

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Morich; Michael A.	Mentor	OH		
Petropoulos; Labros S.	Solon	OH		

US-CL-CURRENT: 324/318

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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9. Document ID: US 5497089 A

L10: Entry 9 of 10 File: USPT Mar 5, 1996

US-PAT-NO: 5497089
DOCUMENT-IDENTIFIER: US 5497089 A

TITLE: Wide aperture gradient set

DATE-ISSUED: March 5, 1996

INVENTOR-INFORMATION:

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Morich; Michael A.	Mentor	OH		
Petropoulos; Labros	Cleveland Heights	OH		

US-CL-CURRENT: 324/318; 324/322

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
Draw Desc | Image

10. Document ID: US 5343148 A

L10: Entry 10 of 10 File: USPT Aug 30, 1994

US-PAT-NO: 5343148
DOCUMENT-IDENTIFIER: US 5343148 A

TITLE: Gradient coil system

DATE-ISSUED: August 30, 1994

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Westphal; Michael	Offenbach			DE
Knuttel; Bertold	Rheinstetten-Morsch			DE
Schmidt; Hartmut	Karlsruhe			DE

US-CL-CURRENT: 324/309; 324/318

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
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Term	Documents
OPEN.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	1909104
OPENS.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	335235
VERTICAL$3	0
VERTICAL.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	1619248
VERTICALA.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	9
VERTICALALY.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	4
VERTICALAND.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	6
VERTICALARM.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	1
VERTICALBAR.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	1
VERTICALD.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	3
VERTICALE.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	15
(L5 AND (OPEN OR VERTICAL$3)).USPT,PGPB,JPAB,EPAB,DWPI,TDBD.	10

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DOCUMENT-IDENTIFIER: US 5708360 A
TITLE: Active shield gradient coil for nuclear magnetic resonance imaging apparatus

Abstract Text (1):
An active shield gradient coil for a nuclear magnetic resonance imaging, formed by a primary coil having a fingerprint shaped current distribution formed on an inner cylinder in which at least a part of current return lines is cut out, a shield coil having a current distribution formed on an outer cylinder, and a bridging connection member for connecting cut out part of the current return lines of the primary coil with the shield coil, in which the current distribution of the shield coil is a composition of a first shield pattern for cancelling out a magnetic field produced by the primary coil outside the outer cylinder and a second shield pattern for cancelling out a magnetic field produced by the bridging connection member outside the outer cylinder. The current distribution of the primary coil can have turn back current paths for leading the cut out part of the current return lines on one end of the gradient coil to another end of the gradient coil.

Brief Summary Text (3):
The present invention relates to a nuclear magnetic resonance imaging apparatus, and more particularly, to a gradient coil for generating a gradient magnetic field in a nuclear magnetic resonance imaging apparatus.

Brief Summary Text (5):
In recent years, among various types of medical diagnostic apparatuses developed, the nuclear magnetic resonance imaging (MRI) apparatus has been studied and developed very actively.

Brief Summary Text (6):
As well known, the nuclear magnetic resonance imaging is a method for imaging microscopic chemical and physical information of matters by utilizing the nuclear magnetic resonance phenomenon in which the energy of a radio frequency magnetic field rotating at a specific frequency can be resonantly absorbed by a group of nuclear spins having unique magnetic moments which are placed in a homogeneous static magnetic field.

Brief Summary Text (7):
This nuclear magnetic resonance imaging has attracted much attention because of its capability for imaging not just physical shape information of a living body at high contrast, but also various other types of functional information such as blood flow information, microscopic magnetic field inhomogeneity information, a diffusion information, and chemical shift information.

Brief Summary Text (8):
In an MRI apparatus for carrying out such a nuclear magnetic resonance imaging, depending on a type of an imaging pulse sequence to be used, numerous different manners of switching of the gradient magnetic fields to be superposed onto the static magnetic field will be required.

Brief Summary Text (9):
Examples of the conventionally known imaging pulse sequence to be used in the nuclear magnetic resonance imaging include the usual imaging sequence such as the spin echo sequence and the field echo sequence, the high speed or ultra high speed imaging sequence such as the high speed spin echo sequence, the high speed field echo sequence, and the echo planar sequence, and the nuclear magnetic resonance angiography sequence for obtaining a distribution or a speed of blood flow in blood

vessels.

Brief Summary Text (10):
Each of these imaging pulse sequences is associated with a characteristic manner of switching of the gradient magnetic fields, and such a switching of the gradient magnetic fields is known to generate eddy currents on thermal shields and a helium container of the superconducting magnet. These eddy currents can affect temporal and spatial characteristics of the gradient magnetic fields to cause a serious degradation of the image quality in the nuclear magnetic resonance images to be obtained such as blurring.

Brief Summary Text (11):
In order to avoid this problem, there has been a proposition of the so called active shield gradient coil (ASGC). This ASGC is a coil in coaxial double cylinder shape which comprises a primary coil for generating the gradient magnetic field in an interior region by means of current distribution on a cylindrical surface, and a shield coil for effectively cancelling out a leaking magnetic field from the primary coil at an exterior region by means of current distribution of a cylindrical surface enclosing the primary coil.

Brief Summary Text (12):
FIG. 1 shows a typical shield pattern for the ASGC proposed by Roemer et al. The ASGC generates the gradient magnetic field in a direction perpendicular to that of the static magnetic field, so that the primary coil and the shield coil have spiral shaped current distributions, and each of the primary coil and the shield coil has a current portion for currents generating a desired gradient magnetic field (referred hereafter as a gradient field generating current) and a current portion for currents to be simply returned (referred hereafter as a current return).

Brief Summary Text (14):
As a solution to this problem, there are conventional propositions for the gradient coil (to be referred hereafter as a simple cutting gradient coil) in which a part or a whole of the current returns of the ASGC is simply cut out and then joined, as disclosed in Japanese Patent Application Laid Open No. 4-144543 (1992) and Wong, Hyde: 11th Annual Meetings of Society of Magnetic Resonance in Medicine, 1992, p. 711. These propositions were aiming at a realization of a high speed gradient magnetic field switching characteristic (or large gradient magnetic field strength) by the reduction of the inductance and the resistance due to the cutting of a part of the current returns.

Brief Summary Text (15):
However, this approach has a problem that the gradient magnetic field linearity (or imaging field of view) and the leaking magnetic field shielding power are considerably degraded because of the simple cutting of the primary coil and the shield coil and the presence of connection wirings between the primary coil and the shield coil. For this reason, the tolerable amount of cutting is limited to a very small amount, and consequently the reduction of the inductance and the resistance is also limited to a very small level.

Brief Summary Text (18):
As a solution to these problems, there is a conventional proposition for an asymmetric gradient coil in which the current turns on one side in an axial direction of one coil are turned back to the current turns on another side. FIG. 2 shows a configuration of the gradient coil for a transverse direction proposed by Roemer, as disclosed in Japanese Patent Application Laid Open No. 5-269099 (1993). In this asymmetric gradient coil, a rate of the imaging field of view with respect to an axial length becomes large, and it becomes possible to obtain a sufficient imaging field of view even in a case in which an axial length of a coil is limited.

Brief Summary Text (20):
As a method for cancelling this torque, there are propositions for the asymmetric gradient coil in which additional current returns for cancellation purpose are incorporated, by AbdulJalil et al. and also by Petrapoulas et al., both in 12th Annual Meetings of Society of Magnetic Resonance in Medicine, 1993. In FIG. 3, this type of the asymmetric gradient coil as proposed by Abduljalil et al. is shown. However, this approach has a problem in that the inductance is even more increased due to the additional current returns.

Brief Summary Text (26):

It is another object the present invention to provide a nuclear magnetic resonance imaging apparatus incorporating the above noted gradient coil.

Brief Summary Text (27):
According to one aspect of the present invention there is provided a gradient coil for a nuclear magnetic resonance imaging, comprising: a primary coil having a fingerprint shaped current distribution formed on an inner cylinder in which at least a part of current return lines is cut out; a shield coil having a current distribution formed on an outer cylinder; and a bridging connection member for connecting cut out part of the current return lines of the primary coil with the shield coil; wherein the current distribution of the shield coil is a composition of a first shield pattern for cancelling out a magnetic field produced by the primary coil outside the outer cylinder and a second shield pattern for cancelling out a magnetic field produced by the bridging connection member outside the outer cylinder.

Brief Summary Text (28):
According to another aspect of the present invention there is provided a gradient coil for a nuclear magnetic resonance imaging, comprising: a primary coil having a fingerprint shaped current distribution formed on an inner cylinder in which at least a part of current return lines is cut out, and turn back current paths for leading the cut out part of the current return lines on one end of the gradient coil to another end of the gradient coil; and a shield coil having a current distribution formed on an outer cylinder for cancelling out a magnetic field produced by the primary coil outside the outer cylinder.

Brief Summary Text (29):
According to another aspect of the present invention there is provided a nuclear magnetic resonance imaging apparatus, comprising: a gradient coil including: a primary coil having a fingerprint shaped current distribution formed on an inner cylinder in which at least a part of current return lines is cut out; a shield coil having a current distribution formed on an outer cylinder; and a bridging connection member for connecting cut out part of the current return lines of the primary coil with the shield coil; wherein the current distribution of the shield coil is a composition of a first shield pattern for cancelling out a magnetic field produced by the primary coil outside the outer cylinder and a second shield pattern for cancelling out a magnetic field produced by the bridging connection member outside the outer cylinder; and imaging means for imaging a body to be examined placed in a homogeneous static magnetic field by applying radio frequency magnetic field onto the body to be examined and operating the gradient coil to apply gradient magnetic fields onto the body to be examined according to a pulse sequence, detecting nuclear magnetic resonance signals emitted from the body to be examined in response to the radio frequency magnetic field and the gradient magnetic fields, and processing the nuclear magnetic resonance signals to construct nuclear magnetic resonance images.

Brief Summary Text (30):
According to another aspect of the present invention there is provided a nuclear magnetic resonance imaging apparatus, comprising: a gradient coil including: a primary coil having a fingerprint shaped current distribution formed on an inner cylinder in which at least a part of current return lines is cut out and turn back current paths for leading the cut out part of the current return lines on one end of the gradient coil to another end of the gradient coil; and; a shield coil having a current distribution formed on an outer cylinder for cancelling out a magnetic field produced by the primary coil outside the outer cylinder; and imaging means for imaging a body to be examined placed in a homogeneous static magnetic field by applying radio frequency magnetic field onto the body to be examined and operating the gradient coil to apply gradient magnetic fields onto the body to be examined according to a pulse sequence, detecting nuclear magnetic resonance signals emitted from the body to be examined in response to the radio frequency magnetic field and the gradient magnetic fields, and processing the nuclear magnetic resonance signals to construct nuclear magnetic resonance images.

Drawing Description Text (6):
FIG. 4 is a block diagram of a nuclear magnetic resonance imaging apparatus suitable for using a gradient coil according to the present invention.

Drawing Description Text (23):
FIG. 18B is a development of a coil pattern for a shield coil in the fourth embodiment of the gradient coil according to the present invention.

Drawing Description Text (25):
FIG. 19B is a development of a coil pattern for a shield coil in the fifth embodiment of the gradient coil according to the present invention.

Drawing Description Text (28):
FIG. 21B is a development of a coil pattern for a shield coil in the sixth embodiment of the gradient coil according to the present invention.

Detailed Description Text (2):
Now, with references to the drawings, various embodiment of a gradient coil for a nuclear magnetic resonance imaging apparatus according to the present invention will be described in detail.

Detailed Description Text (3):
First, for all the embodiments of the gradient coil to be described in detail below, a suitable nuclear magnetic resonance imaging apparatus has a configuration as shown in FIG. 4.

Detailed Description Text (4):
This nuclear magnetic resonance imaging apparatus of FIG. 4 comprises: a main magnet 101 for generating a static magnetic field; a main magnet power source 102 for driving the main magnet 101; primary gradient coils 103 for generating gradient magnetic fields; shield coils 104 provided around the primary gradient coils 103; a gradient coil power source 105 for driving the primary gradient coils 103 and the shield coils 104; an eddy current compensation circuit 107 for adjusting inputs to the gradient coil power source 105 so as to compensate the effect due to the eddy currents; shim coils 108 for adjusting the homogeneity of the static magnetic field; shim coil power source 109 for driving the shim coils 108; a probe 111 for applying radio frequency (RF) pulses to a body to be examined 106 and receiving nuclear magnetic resonance (NMR) signals from the body to be examined 106; an RF shielding 112 provided between the primary gradient coils 103 and the probe 111; a transmitter unit 110 for driving the probe 111 to transmit the desired RF pulses; and a receiver unit 113 for detecting the NMR signals received by the probe 111.

Detailed Description Text (5):
In addition, this apparatus of FIG. 4 further comprises: a data acquisition unit 114 for acquiring and A/D converting the NMR signals detected by the receiver unit 113; a data processing unit 115 for data processing the A/D converted NMR signals to reconstruct the desired NMR image; a display unit 118 for displaying the NMR image reconstructed by the data processing unit 115; a system controller 116 for controlling the operations of the main magnet power source 102, the gradient coil power source 105, the eddy current compensation circuit 107, the transmitter unit 110, the receiver unit 113, the data acquisition unit 114, and the data processing unit 115, so as to realize the desired imaging pulse sequence; and a console 117 from which an operator enters various control commands to the system controller 116 and the data processing unit 115.

Detailed Description Text (6):
Here, the main magnet 101 is driven by the main magnet power source 102 while the primary gradient coils 103 and the shield coils 104 are driven by the gradient coil power source 105 such that a uniform static magnetic field and the gradient magnetic fields having linear gradients in three mutually orthogonal directions are applied onto the body to be examined 106. The primary gradient coils 103 and the shield coils 104 may be connected in series and driven by the common gradient coil power source 105, or upper, lower, right, and left coil elements may be separately connected with a plurality of gradient coil power sources 105 and separately driven by the respective gradient coil power sources 105. The input signals to be given to the gradient coil power source 105 for compensating eddy current magnetic field time response are generated at the eddy current compensation circuit 107.

Detailed Description Text (7):
In this apparatus of FIG. 4, the body to be examined 106 is placed inside the static magnetic field generated by the main magnet 101, and three orthogonal gradient magnetic fields generated by the primary gradient coils 103 are superposed onto the static magnetic field while the RF pulses are applied by the probe 111, according to the desired imaging pulse sequence. Then, the NMR signals emitted from the patient 106 in response to the application of the RF pulses are received by the probe 111. Here, the common probe 111 may be used for the transmission of the RF pulses and the

reception of the NMR signals, or separate probes 111 may be provided for the transmission of the RF pulses and the reception of the NMR signals.

Detailed Description Text (8):
The NMR signals received by the probe 111 is detected at the receiver unit 113, A/D converted at the data acquisition unit 114, and sent to the data processing unit 115 which reconstructs the desired NMR images by using appropriate data processing operations such as the Fourier transformation. The reconstructed NMR images are then displayed at the display unit 118.

Detailed Description Text (11):
The gradient coil of FIG. 5A comprises a primary coil 11, a shield coil 12, and a plurality of connecting faces 13 for connecting the primary coil 11 and the shield coil 12, where the primary coil 11 has a current distribution mainly composed of the gradient field generating currents (i.e., current portions for currents generating a desired gradient magnetic field) and the shield coil 12 has a current distribution for effectively cancelling out a magnetic field produced by the primary coil 11 and the connecting faces 13 at an exterior region. Each connecting face 13 includes current components 14 (to be referred hereafter as bridging lines) for connecting the primary coil 11 and the shield coil 12 and current components 15 (to be referred hereafter as remaining current returns) for connecting ends of the horseshoe shaped current patterns on the primary coil 11.

Detailed Description Text (12):
The current distributions of these primary coil 11, shield coil 12, and connecting faces 13 are actually realized by conductors or copper wires in rectangular cross sections which are manufactured into the current distribution patterns by etching, etc.

Detailed Description Text (13):
Here, the shield coil 12 in the gradient coil of FIG. 5A is shaped in a pattern for shielding the magnetic field produced by the current distributions of the primary coil 11 and the connecting faces 13 by means of the designing algorithm to be described in detail below, in contrast to the conventional simple cutting gradient coil of FIG. 5B which is not shaped in a pattern for shielding the leaking magnetic field from the primary coil and the connection wirings.

Detailed Description Text (14):
In addition, the gradient coil of FIG. 5A is shaped in a pattern for securing the gradient magnetic field linearity when all the current distributions are present on the primary coil 11, the shield coil 12, and the connecting faces 13 by means of the designing algorithm to be described in detail below, in contrast to the conventional simple cutting gradient coil of FIG. 5B in which the gradient magnetic field linearity is degraded due to the simple cutting.

Detailed Description Text (15):
Now, the gradient coil of this first embodiment can be formed in a structure shown in FIG. 6, which shows a cross sectional view in an axial direction of a layer structure for both X and Y gradient coils in which the current distributions of the connecting faces 13 are formed by connecting the conductors manufactured into the current distribution patterns by etching, etc. to the primary coil conductor and the shield coil conductor by means of screws 134.

Detailed Description Text (16):
More specifically, in FIG. 6, a primary coil conductor layer 135 and a shield coil conductor layer 137 for the X coil are connected by a bridging line 131, a primary coil conductor layer 136 and a shield coil conductor layer 138 for the Y coil are connected by a bridging line 132. Here, a coil pair of the primary coil and the shield coil for the X coil and a coil pair of the primary coil and the shield coil for the Y coil are formed as nested coil pairs. That is, the primary coil 135 for the X coil is located at a position inside the primary coil 136 for the Y coil in a radial direction, and the shield coil 137 for the X coil is located at a position outside the shield coil 138 for the Y coil in a radial direction. It is to be noted that the layer order for the X coil and the Y coil shown in FIG. 6 may be reversed if desired.

Detailed Description Text (20):
First, at the step S21, the structures of the primary coil and the shield coil such as their radii are determined, and the high efficiency current distribution for the

primary coil which is mainly composed of the gradient field generating currents is set up.

Detailed Description Text (21):
Here, the magnetic field B produced by the current distribution j can be expressed in general by the following equation (1), according to Carlson, et al.: Magnetic Resonance in Medicine, 26, p. 191, 1992, for example. ##EQU1## where: .mu..sub.0 : a magnetic susceptibility of vacuum,

Detailed Description Text (26):
The primary coil current distribution j.sup.p and the shield coil current distribution j.sup.s can be expressed in general by the following equations (2) and (3).

Detailed Description Text (29):
R': a radius of the shield coil

Detailed Description Text (35):
Next, at the step S22, the current distribution for the shield coil is determined to obtain a shield pattern corresponding to the primary coil, for desired imaging field of view and shielding power. Here, the current distribution can be determined from the condition that the magnetic field becomes zero outside the shield coil. Here, the procedure for determining the current distribution for the shield coil will be briefly outlined without going into too much details that can be found elsewhere.

Detailed Description Text (38):
In this manner, as indicated in a part (a) of FIG. 9, the shield pattern 42 corresponding to the primary coil portion 41 alone is obtained. This shield pattern 42 is then subjected to the pattern cutting.

Detailed Description Text (39):
Next, at the step S23, the connecting current distribution is determined by cutting the return portions in the shield pattern. Here, as indicated in a part (b) of FIG. 9, a part or a whole of the current returns is cut out, and the current turns of the shield coil and the current turns of the primary coil are connected by the bridging lines 43.

Detailed Description Text (40):
Here, in general, a number of current turns in the shield coil is smaller than a number the primary coil, in the primary coil, so that there remains some current turns (referred hereafter as remaining turns) in the primary coil which are not connected with the current turns of the shield coil. As indicated in a part (c) of FIG. 9, these remaining turns are connected by the remaining current returns 45 on the primary coil cylinder surface or the connecting face for connecting the primary coil and the shield coil.

Detailed Description Text (44):
Next, at the step S25, as indicated in a part (d) of FIG. 9, a composed shield pattern 47 is obtained by superposing the shield pattern 42 corresponding to the primary coil determined at the step S21 and the shield patterns 44 and 46 corresponding to the connecting current distribution determined at the step S24.

Detailed Description Text (46):
Next, at the step S27, the primary coil 41 and the corresponding shield pattern 42 for cancelling the gradient magnetic field degraded component are determined for a desired imaging field of view. This can be done by the similar operation as in the above steps S21 and S22, except that it is necessary to add an opposite sign for the gradient magnetic field degraded component to the above described equations (8) and (9).

Detailed Description Text (47):
Then, after pattern cutting the obtained shield pattern 42, the operation returns to the above described step S23. Thereafter, the series of operation in the loop of the steps S23 to S27 is repeated until the optimized coil current patterns for the primary coil, the shield pattern, and the connecting current distribution are finally obtained at the step S28.

Detailed Description Text (48):
In this manner, it becomes possible to make the leaking magnetic field produced by

the current distributions of the primary coil and the connecting face to be nearly zero at an exterior region of the shield coil, while generating the gradient magnetic field which has a good linearity in the desired imaging field of view.

Detailed Description Text (50):
First, at the step S31, the conventional current distribution for the primary coil is set up according to the same current distribution base function of FIG. 8B as the conventional ASGC in which the current sum is equal to 0. Then, at the step S32, the current distribution for the shield coil is determined to obtain a shield pattern corresponding to the primary coil, for desired imaging field of view and shielding power, as in indicated in a part (a) of FIG. 11. Then, at the step S33, after the pattern cutting of the primary coil and the shield coil, the connection wirings are determined, and the designing is finished. Here, as indicated in parts (b) and (c) of FIG. 11, no consideration is given to the gradient magnetic field linearity (or the imaging field of view) and the shielding power, so that the degradation due to the presence of the bridging lines and the remaining current returns is left in the designed gradient coil.

Detailed Description Text (51):
Namely, the conventional simple cutting gradient coil achieves a higher field generation efficiency by sacrificing the imaging field of view and the shielding power, in contrast to the present invention in which the imaging field of view and the shielding power comparable to the conventional ASGC are maintained while generating the gradient magnetic field at a high field generation efficiency.

Detailed Description Text (52):
Thus, the gradient coil of this first embodiment is constructed from a primary coil current distribution which is mainly composed of the gradient field generating currents while maintaining the imaging field of view and the shielding power comparable to the conventional ASGC, so that a number of current returns is reduced, and therefore the inductance, the coil resistance, the local heat generation, and the driving power can be reduced compared with the conventional ASGC for generating the same gradient magnetic field strength while the axial length of the gradient coil can be made shorter. With the shorter axial length, a noise and a vibration at a time of the gradient magnetic field switching can be reduced, and a weight and a manufacturing cost of the gradient coil can also be reduced.

Detailed Description Text (57):
In the conventional cylindrical gradient coil for the head portion imaging, the axial length of the gradient coil has been limited by the shoulders of a patient 54, but in the gradient coil of this second embodiment, the sectional cuttings 52 can provide extra rooms for the shoulders of a patient 54 so that the gradient coil with a longer axial length can be realized.

Detailed Description Text (58):
According to Carlson, et al.: Magnetic Resonance in Medicine, 26, p. 191, 1992, already mentioned above for example, the gradient coil with a longer axial length tends to have a higher field generation efficiency, so that it is possible for the gradient coil of this second embodiment to realize a higher field generation efficiency.

Detailed Description Text (64):
The gradient coil of FIG. 14 comprises a primary coil 91, a shield coil 92, and a connecting face 93 for connecting the primary coil 91 and the shield coil 92, where the primary coil 91 has a current distribution mainly composed of the gradient field generating currents and the shield coil 92 has a current distribution for effectively cancelling out a magnetic field produced by the primary coil 91 and the connecting face 93 at an exterior region. The connecting face 93 includes the bridging lines 94 for connecting the primary coil 91 and the shield coil 92 and the remaining current returns 95 for connecting ends of the horseshoe shaped current patterns on the primary coil 91.

Detailed Description Text (66):
The current distributions of these primary coil 91, shield coil 92, and connecting face 93 are actually realized by conductors or copper wires in rectangular cross sections which are manufactured into the current distribution patterns by the etching, etc.

Detailed Description Text (67):

Here, the shield coil 99 in the gradient coil of FIG. 14 is shaped in a pattern for shielding the magnetic field produced by the current distributions of the primary coil 91 and the connecting face 93 by means of the designing algorithm similar to that used in the first embodiment described above, which will now be described.

Detailed Description Text (69):
First, at the step S21, the structures of the primary coil and the shield coil such as their radii are determined, and the high efficiency current distribution for the primary coil which is mainly composed of the gradient field generating currents is set up.

Detailed Description Text (70):
Here, the magnetic field B produced by the current distribution j can be expressed in general by the above equation (1). Also, the primary coil current distribution j.sup.p and the shield coil current distribution j.sup.s can be expressed in general by the above equations (2) and (3).

Detailed Description Text (73):
Next, at the step S22, the current distribution for the shield coil is determined to obtain a shield pattern corresponding to the primary coil, for desired imaging field of view and shielding power. Here, the procedure for determining the current distribution for the shield coil is substantially the same as that described above for the first embodiment in conjunction with the above equations (6) to (11). In this manner, as indicated in a part (a) of FIG. 15, the shield pattern 122 corresponding to the primary coil portion 121 alone is obtained. This shield pattern 122 is then subjected to the pattern cutting. Next, at the step S23, the connecting current distribution is determined by cutting the return portions in the shield pattern. Here, as indicated in a part (b) of FIG. 11, a part or a whole of the current returns is cut out, and the current turns of the shield coil and the current turns of the primary coil are connected by the bridging lines 124.

Detailed Description Text (74):
Also, in general, a number of current turns in the shield coil is smaller than a number of current turns in the primary coil, so that there remains the remaining turns in the primary coil. As indicated in a part (c) of FIG. 11, these remaining turns are connected by the remaining current returns 126 on the primary coil cylinder surface or the connecting face for connecting the primary coil and the shield coil.

Detailed Description Text (76):
Next, at the step S25, as indicated in a part (d) of FIG. 15, a composed shield pattern 118 is obtained by superposing the shield pattern 122 corresponding to the primary coil determined at the step S21 and the shield patterns 125 and 127 corresponding to the connecting current distribution determined at the step S24.

Detailed Description Text (78):
Next, at the step S27, the primary coil 121 and the corresponding shield pattern 122 for cancelling the gradient magnetic field degraded component are determined for a desired imaging field of view, just as in the first embodiment.

Detailed Description Text (79):
Then, after pattern cutting the obtained shield pattern 122, the operation returns to the above described step S23. Thereafter, the series of operation in the loop of the steps S23 to S27 is repeated until the optimized coil current patterns for the primary coil, the shield pattern, and the connecting current distribution are finally obtained at the step S28.

Detailed Description Text (80):
In this manner, it becomes possible to make the leaking magnetic field produced by the current distributions of the primary coil and the connecting face to be nearly zero at an exterior region of the shield coil, while generating the gradient magnetic field which has a good linearity in the desired imaging field of view. Here, due to the magnetic field cancellation effect by the shield coil, a number of turns in the primary coil is greater than that in a non-shielding type gradient coil, but the increase of the inductance is suppressed by the cutting of the current returns.

Detailed Description Text (81):
In addition, as the directions of the currents are opposite in the primary coil and

the shield coil, the Lorentz forces exerted on the current turns cancel each other, so that it is possible to reduce the torque exerted on the gradient coil as a whole.

Detailed Description Text (82):
Here, the Lorentz forces exerted on the bridging lines appear as shown in FIG. 16, which shows a view of the gradient coil in a direction of the static magnetic field, where a circle 57 represents the primary coil and a circle 58 represents the shield coil. In this case, the Lorentz force 55 exerted on each bridging line 56 is perpendicular to a direction of currents in the bridging line 56, and as a whole, a force in a negative x-direction is exerted. On the other hand, the Lorentz forces exerted on the remaining current returns appear as shown in FIG. 17, which also shows a view of the gradient coil in a direction of the static magnetic field, where a circle 63 represents the primary coil and a circle 64 represents the shield coil. In this case, the Lorentz force 61 exerted on each remaining current return 62 is perpendicular to a direction of currents in each remaining current return 62, and as a whole, a force in a positive x-direction is exerted. Thus, the Lorentz forces exerted on the bridging lines and the remaining current returns cancel each other, and therefore the generated torque is small.

Detailed Description Text (86):
In this fourth embodiment, the gradient coil comprises the primary coil 71 with a current turn configuration as shown in FIG. 18A and the shield coil 72 with a current turn configuration as shown in FIG. 18B, where FIGS. 18A and 18B show a 1/2 region of the respective current turn configurations developed into a plane.

Detailed Description Text (88):
Here, the z-direction current path 74 generates no z-direction magnetic field component, so that it does not affect the gradient magnetic field linearity at all, and no Lorentz force is exerted thereon. In other words, each of the primary coil and the shield coil of this fourth embodiment is constructed from the fingerprint shaped current distribution which generates the gradient magnetic field, and the z-direction current paths which do not generate the gradient magnetic field.

Detailed Description Text (90):
In this fifth embodiment, the gradient coil comprises the primary coil with a current turn configuration as shown In FIG. 19A and the shield coil with a current turn configuration as shown in FIG. 19B, where FIGS. 19A and 19B show a 1/2 region of the respective current turn configurations developed into a plane.

Detailed Description Text (91):
In this fifth embodiment, an end point of each turn on one side end of each coil is turned back to the other side end of each coil, by means of the z-direction current path 83, just as in the fourth embodiment described above. Then, a part of the current turns of the primary coil similar to that of the fourth embodiment described above are connected with the current turns of the shield coil similar to that of the fourth embodiment described above by the bridging lines 81, while the rest of the current turns of the primary coil are connected by the remaining current returns 82, and the composed shield pattern is formed to cancel out the leaking magnetic field due to the primary coil, the bridging lines 81 and the remaining current returns 82.

Detailed Description Text (92):
In this case, because of the use of the bridging lines 81, a number of remaining current returns 82 in the primary coil and the shield coil can be reduced, so that the inductance of the gradient coil can be reduced compared with the fourth embodiment described above.

Detailed Description Text (94):
More specifically, in FIG. 20, a bridging line 142 connects a primary coil fingerprint shaped current distribution layer 143 and a shield coil fingerprint shaped current distribution layer 144 for the X coil, while a bridging line 145 connects a primary coil z-direction current path layer 146 and a shield coil z-direction current path layer 147 for the X coil. Similarly, a bridging line 148 connects a primary coil fingerprint shaped current distribution layer 149 and a shield coil fingerprint shaped current distribution layer 150 for the Y coil, while a bridging line 151 connects a primary coil z-direction current path layer 152 and a shield coil z-direction current path layer 153 for the Y coil. The fingerprint shaped current distribution layer and the z-direction current path layer for each of

the primary coil and the shield coil are connected together in a region 154, current turn by current turn.

Detailed Description Text (96):
In FIG. 20, a coil assembly of the fingerprint shaped current distribution layers for the primary coil and the shield coil and the bridging line connecting them is formed to enclose a coil assembly of the z-direction current path layers for the primary coil and the shield coil and the bridging line connecting them, but it is also possible to reverse this layer order such that the latter encloses the former.

Detailed Description Text (98):
In this sixth embodiment, the gradient coil comprises the primary coil with a current turn configuration as shown in FIG. 21A and the shield coil with a current turn configuration as shown in FIG. 21B, where FIGS. 21A and 21B show a 1/2 region of the respective current turn configurations developed into a plane.

Detailed Description Text (99):
In this sixth embodiment, the composed shield pattern is formed to cancel out the leaking magnetic field due not only to the primary coil, the bridging lines, and the remaining current returns, but also to the z-direction current paths in the primary coil as well.

Detailed Description Text (101):
Thus, the z-direction current paths can be handled just like the bridging lines and the remaining current returns, by obtaining the shield pattern corresponding to the z-direction current paths in the primary coil at the step S24 of FIG. 7, and obtaining the gradient magnetic field degraded component due to the cutting of the composed shield pattern at the step S26 of FIG. 7, so as to cancel out the gradient magnetic field degradation in the same manner as described above for the first embodiment.

Detailed Description Text (102):
In this manner, it is possible to realize the gradient coil with a higher shielding power compared with the fifth embodiment described above.

Detailed Description Text (103):
Now, the gradient coil of this sixth embodiment can be formed in a structure shown in FIG. 22, which shows a cross sectional view in an axial direction of a layer structure for both X and Y gradient coils. Here, each of the X coil and the Y coil is formed by three layers of conductors, with an insulative layer inserted between each adjacent conductors. In each of the X coil and the Y coil, the primary coil has the fingerprint shaped current distribution layer and the z-direction current path layer, while the shield coil has a single shield coil conductor layer.

Detailed Description Text (104):
More specifically, in FIG. 22, a primary coil fingerprint shaped current distribution layer 156 and a primary coil z-direction current path layer 159 for the X coil are connected with a shield coil conductor layer 157 for the X coil by bridge lines 155 and 158, while a primary coil fingerprint shaped current distribution layer 161 and a primary coil z-direction current path layer 164 for the Y coil are connected with a shield coil conductor layer 162 for the Y coil by bridge lines 160 and 163.

Detailed Description Text (107):
Thus, the gradient coil of this sixth embodiment has a higher shielding power compared with the fifth embodiment described above, an equivalent field generation efficiency as the third embodiment described above, and a smaller local heat generation compared with the third embodiment described above.

Other Reference Publication (1):
Wong et al., Society of Magnetic Resonance in Medicine, Book of Abstracts. vol. 1, "Short Cylindrical Tranverse Gradient Ciols Using Remote Current Return", p. 583, (1992).

Other Reference Publication (2):
Abduljail et al., Proceddings of the Society of Magnetic Resonance in Medicine, vol. 3, "Torque Compensated Asymmetric Gradient Coils for EPI", p. 1306, (1993).

Other Reference Publication (3):

Abduljail et al., Magnetic Resonance in Medicine, vol. 31, "Torque Free Asymmetric Gradient Coils for Echo Planar Imaging", pp. 450-453, (1994).

Other Reference Publication (4):
Carlson et al., "Design and Evaluation of Shielded Gradient Coils", Magnetic Resonance In Medicine No. 26, (1992), pp. 191-206.

Other Reference Publication (5):
Myers et al., Society of Magnetic Resonance in Medicine, Book of Abstracts, vol. 2, "Highly Linear Asymmetric Transverse Gradient Coil Design for Head Imaging", p. 711, (1991).

CLAIMS:

1. A gradient coil for nuclear magnetic resonance imaging, comprising:

a primary coil having a fingerprint shaped current distribution formed on an inner cylinder in which at least a part of current return lines is cut out to form a cut out part with broken current return lines;

a shield coil having a current distribution formed on an outer cylinder; and

a bridging connection member for connecting the cut out part of the current return lines of the primary coil with the shield coil;

wherein the current distribution of the shield coil is a composition of a first shield pattern for cancelling out a magnetic field produced by the primary coil outside the outer cylinder and a second shield pattern for cancelling out a magnetic field produced by the bridging connection member outside the outer cylinder.

2. The gradient coil of claim 1, further comprising a return connection member for connecting remaining current turns of the primary coil, the remaining current turns being current turns in the cut out part of the current return lines of the primary coil which are not connected with the shield coil by the bridging connection member, and wherein the current distribution of the shield coil is a composition of the first shield pattern, the second shield pattern, and a third shield pattern for cancelling out a magnetic field produced by the return connection member outside the outer cylinder.

4. The gradient coil of claim 1, wherein the fingerprint shaped current distribution of the primary coil, the current distribution of the shield coil, and a current distribution of the bridging connection member are set in a current distribution pattern for realizing a desired gradient magnetic field linearity in a desired imaging field of view within the inner cylinder.

6. The gradient coil of claim 1, wherein a first channel coil assembly formed by a first channel part of the primary coil, a first channel part of the shield coil, and a first channel part of the bridging connection member is enclosed within a second channel coil assembly formed by a second channel part of the primary coil, a second channel part of the shield coil, and a second channel part of the bridging connection member.

10. The gradient coil of claim 7, wherein the bridging connection member connects the cut out part of the current return lines and the turn back current paths of the primary coil with the shield coil.

12. The gradient coil of claim 7, wherein the shield coil has a current distribution in which at least a part of shield coil current return lines is cut out to form a cut out part with broken shield coil current return lines and shield coil turn back current paths for leading the cut out part of the shield coil current return lines on one end of the gradient coil to said another end of the gradient coil.

13. The gradient coil of claim 12, wherein the shield coil turn back current paths are substantially parallel to an axial direction of the gradient coil.

14. The gradient coil of claim 12, further comprising a shield coil return connection member for connecting the cut out part of the shield coil current return lines together at said another end of the gradient coil.

15. The gradient coil of claim 12, wherein the bridging connection member connects the cut out part of the current return lines and the turn back current paths of the primary coil with the shield coil current return lines and the shield coil turn back current paths.

16. The gradient coil of claim 12, wherein the current distribution and the turn back current paths of the shield coil are formed on separate layers.

17. A gradient coil for nuclear magnetic resonance imaging, comprising:

a primary coil having a fingerprint shaped current distribution formed on an inner cylinder in which at least a part of current return lines is cut out to form a cut out part with broken current return lines, and turn back current paths for leading the cut out part of the current return lines on one end of the gradient coil to another end of the gradient coil; and

a shield coil having a current distribution formed on an outer cylinder for cancelling out a magnetic field produced by the primary coil outside the outer cylinder.

20. The gradient coil of claim 17, further comprising a bridging connection member for connecting the cut out part of the current return lines and the turn back current paths of the primary coil with the shield coil.

22. The gradient coil of claim 17, wherein the shield coil has a current distribution in which at least a part of shield coil current return lines is cut out to form a cut out part with broken shield coil current return lines and shield coil turn back current paths for leading the cut out part of the shield coil current return lines on one end of the gradient coil to said another end of the gradient coil.

23. The gradient coil of claim 22, wherein the shield coil turn back current paths are substantially parallel to an axial direction of the gradient coil.

24. The gradient coil of claim 22, further comprising a shield coil return connection member for connecting the cut out part of the shield coil current return lines together at said another end of the gradient coil.

25. The gradient coil of claim 22, further comprising a bridging connection member for connecting the cut out part of the current return lines and the turn back current paths of the primary coil with the shield coil current return lines and the shield coil turn back current paths.

26. The gradient coil of claim 22, wherein the current distribution and the turn back current paths of the shield coil are formed on separate layers.

27. The gradient coil of claim 17, wherein a first channel coil assembly formed by a first channel part of the primary coil and a first channel part of the shield coil is enclosed within a second channel coil assembly formed by a second channel part of the primary coil and a second channel part of the shield coil.

28. A nuclear magnetic resonance imaging apparatus, comprising:

a gradient coil including

a primary coil having a fingerprint shaped current distribution formed on an inner cylinder in which at least a part of current return lines is cut out to form a cut out part with broken current return lines;

a shield coil having a current distribution formed on an outer cylinder; and

a bridging connection member for connecting the cut out part of the current return lines of the primary coil with the shield coil;

wherein the current distribution of the shield coil is a composition of a first shield pattern for cancelling out a magnetic field produced by the primary coil outside the outer cylinder and a second shield pattern for cancelling out a magnetic field produced by the bridging connection member outside the outer cylinder; and

imaging means for imaging a body to be examined placed in a homogeneous static magnetic field by applying a radio frequency magnetic field onto the body to be examined and operating the gradient coil to apply gradient magnetic fields onto the body to be examined according to a pulse sequence, detecting nuclear magnetic resonance signals emitted from the body to be examined in response to the radio frequency magnetic field and the gradient magnetic fields, and processing the nuclear magnetic resonance signals to construct nuclear magnetic resonance images.

29. The apparatus of claim 28, further comprising a return connection member for connecting remaining current turns of the primary coil, the remaining current turns being current turns in the cut out part of the current return lines of the primary coil which are not connected with the shield coil by the bridging connection member, and wherein the current distribution of the shield coil is a composition of the first shield pattern, the second shield pattern, and a third shield pattern for cancelling out a magnetic field produced by the return connection member outside the outer cylinder.

31. The apparatus of claim 28, wherein the fingerprint shaped current distribution of the primary coil, the current distribution of the shield coil, and a current distribution of the bridging connection member are set in a current distribution pattern for realizing a desired gradient magnetic field linearity in a desired imaging field of view within the inner cylinder.

33. The apparatus of claim 28, wherein a first channel coil assembly formed by a first channel part of the primary coil, a first channel part of the shield coil, and a first channel part of the bridging connection member is enclosed within a second channel coil assembly formed by a second channel part of the primary coil, a second channel part of the shield coil, and a second channel part of the bridging connection member.

37. The apparatus of claim 34, wherein the bridging connection member connects the cut out part of the current return lines and the turn back current paths of the primary coil with the shield coil.

39. The apparatus of claim 34, wherein the shield coil has a current distribution in which at least a part of shield coil current return lines is cut out to form a cut out part with broken shield coil current return lines and shield coil turn back current paths for leading the cut out part of the shield coil current return lines on one end of the gradient coil to said another end of the gradient coil.

40. The apparatus of claim 39, wherein the shield coil turn back current paths are substantially parallel to an axial direction of the gradient coil.

41. The apparatus of claim 39, further comprising a shield coil return connection member for connecting the cut out part of the shield coil current return lines together at said another end of the gradient coil.

42. The apparatus of claim 39, wherein the bridging connection member connects the cut out part of the current return lines and the turn back current paths of the primary coil with the shield coil current return lines and the shield coil turn back current paths.

43. The apparatus of claim 39, wherein the current distribution and the turn back current paths of the shield coil are formed on separate layers.

44. A nuclear magnetic resonance imaging apparatus, comprising:

a gradient coil including

a primary coil having a fingerprint shaped current distribution formed on an inner cylinder in which at least a part of current return lines is cut out to form a cut out part with broken current return lines, and turn back current paths for leading the cut out part of the current return lines on one end of the gradient coil to another end of the gradient coil; and

a shield coil having a current distribution formed on an outer cylinder for cancelling out a magnetic field produced by the primary coil outside the outer cylinder; and

imaging means for imaging a body to be examined placed in a homogeneous static magnetic field by applying a radio frequency magnetic field onto the body to be examined and operating the gradient coil to apply gradient magnetic fields onto the body to be examined according to a pulse sequence, detecting nuclear magnetic resonance signals emitted from the body to be examined in response to the radio frequency magnetic field and the gradient magnetic fields, and processing the nuclear magnetic resonance signals to construct nuclear magnetic resonance images.

47. The apparatus of claim 44, further comprising a bridging connection member for connecting the cut out part of the current return lines and the turn back current paths of the primary coil with the shield coil.

49. The apparatus of claim 44, wherein the shield coil has a current distribution in which at least a part of shield coil current return lines is cut out to form a cut out part with broken shield coil current return lines and shield coil turn back current paths for leading the cut out part of the shield coil current return lines on one end of the gradient coil to said another end of the gradient coil.

50. The apparatus of claim 49, wherein the shield coil turn back current paths are substantially parallel to an axial direction of the gradient coil.

51. The apparatus of claim 49, further comprising a shield coil return connection member for connecting the cut out part of the shield coil current return lines together at said another end of the gradient coil.

52. The apparatus of claim 49, further comprising a bridging connection member for connecting the cut out part of the current return lines and the turn back current paths of the primary coil with the shield coil current return lines and the shield coil turn back current paths.

53. The apparatus of claim 49, wherein the current distribution and the turn back current paths of the shield coil are formed on separate layers.

54. The apparatus of claim 44, wherein a first channel coil assembly formed by a first channel part of the primary coil and a first channel part of the shield coil is enclosed within a second channel coil assembly formed by a second channel part of the primary coil and a second channel part of the shield coil.